Proceedings of the 28th Conference of the International Society of Biomechanics in Sports

July 19-23, 2010
Northern Michigan University
Marquette, Michigan USA

Edited by
Randall Jensen, William Ebben, Erich Petushek, Chris Richter, & Karen Roemer

Department of Health Physical Education and Recreation
College of Professional Studies
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The 28th conference of the International Society of Biomechanics in Sports (ISBS)is the official annual meeting of the ISBS whose primary purposes are:

- To provide a forum for the exchange of ideas for sports biomechanists, researchers, coaches, and teachers.
- To bridge the gap between researchers and practitioners.
- To gather and disseminate information and materials on biomechanics in sports

These proceedings were presented at Northern Michigan University in Marquette, MI USA from July 19th to 23rd, 2010. They include eight keynote papers, 89 Oral papers, 106 poster papers, 8 papers from a thematic session on Paralympics, and papers from 6 applied sessions. The conference proceedings present up to date research on sports biomechanics and their applications of researchers from 28 countries around the world.

Authors submitted 250 papers online which were blind reviewed by two experts in the field for the ISBS 2010 Scientific Committee. Authors were required to satisfy requirements for scientific merit and relevance to sport biomechanics, had to respond to reviewers' comments and make requested changes to the approval of the ISBS 2010 Editorial Board. 37 papers were rejected or withdrawn by the authors during the review process. Accepted papers were edited and formatted to ensure they complied with ISBS guidelines. Papers are published as either a four-page paper or a two-page structured abstract.

The proceedings also include 15 applied session papers from the areas of Nordic skiing, Olympic weightlifting, motion analysis, feedback to athletes, writing and reviewing scientific papers, and gymnastics. These sessions allowed practitioners of coaching and teaching to obtain cutting edge information on topics of interest to them from experts in the field.

These proceedings would not have been possible without the assistance of the ISBS 2010 Keynote Speakers, the ISBS 2010 Scientific Committee, and the Applied Session Chairs. Their hard work ensured a high quality that should be evident to the reader of the included papers. We hope that you find the papers useful in your work, be it research, coaching or teaching.

Randall Jensen, William Ebben, Erich Petushek, Chris Richter, & Karen Roemer

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Oral Session 11 – Balance

Oral Session 12 - NIA Kinematics / Kinetics

Oral Session 13 – Throwing

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Poster Session 1

Poster Session 2

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Poster Session 4

ISBS 2010

Keynotes

BIOMECHANICAL MODELING APPLIED TO HUMAN MOVEMENT ANALYSIS

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Biomechanical modelling can be a helpful instrument in human movement analysis. Here, the focus will be set on analyzing human movement using inverse dynamics. Dealing with inverse problems, one important issue is the transfer of a real motion to the model and how to find an optimal description of the movement. Furthermore, joint models, the description of joint kinematics, as well as the individual adaptability of those models will influence the results of human movement analysis.

KEYWORDS: inverse dynamics, biomechanical modelling, motion analysis

INTRODUCTION: One aim of sport biomechanics is to analyze movement techniques during competition or training exercises. This results in knowledge about efficiency of certain techniques and provides a basis for future recommendations for training exercises. This paper presents applications of biomechanical modelling for human movement analysis.

ANALYSIS OF VOLLEYBALL SPIKES: In order to understand how specific movements are performed, it is important to describe the specific movement under consideration with high accuracy. Assets and drawbacks of these movements can only be understood, if it is possible to describe movements with close-to-reality and artefact-poor time histories for the inner co-ordinates.

Roemer et al. 2007 used a 3D man model to reconstruct volleyball spikes performed during European League Games in 2006. Boundary conditions for the analysis were: diagonal performed spikes from position four, flight angle of the ball after impact from 110° to 145° to the net, and the step close technique (Coutts, 1982). Four digital cameras (3x Basler, 1x Vosskühler) with a frame rate of 100 Hz were used. This setup was calibrated by 16 landmarks with known coordinates. Seventeen body points were digitized per frame and the multi body system (MBS) man model DYNAMICUS was used to quantify the time histories of these joint angles. Thirteen joints were considered and the total degrees of freedom for this model was 22. Using the Dynamic Tracking method (Roemer et al., 2001), model-fixed points were defined corresponding to the digitized body points of the subject. In these points the man model was connected viscoelastically to the reference points. This leads to dynamic adjustments of the man model with the moving reference point cloud. Due to the linear elasticity used to connect the markers, this approach is equivalent to linear filtering. Thus, the movement of the real volleyball spiker was transferred to the man model and the time histories of all major joints were quantified.

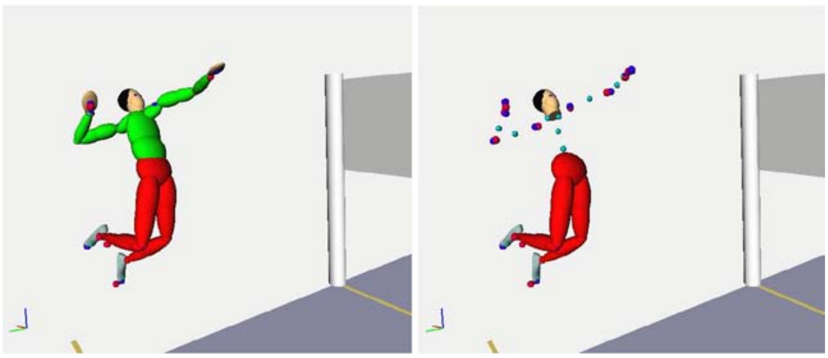

Figure 1: Man model in marker cloud

The investigated movements occurred in different game situations and different sets, but individual coordination techniques within the shoulder joint could be identified for this movement (Roemer et al. 2007).
Roemer et al (2008) analyzed more players and the shoulder kinematics were described in greater detail; therefore the kinematic description of the shoulder movement applied quaternions and the axis-angle approach to avoid the gimbal lock. The orientation of the resulting axis of rotation in the shoulder joint and the rotational angle were calculated. Additionally the 3D coordinates of the elbow movement around the shoulder and the internal and external rotation were investigated. The results show that specific movement strategies for the humerus could be detected using these methods.

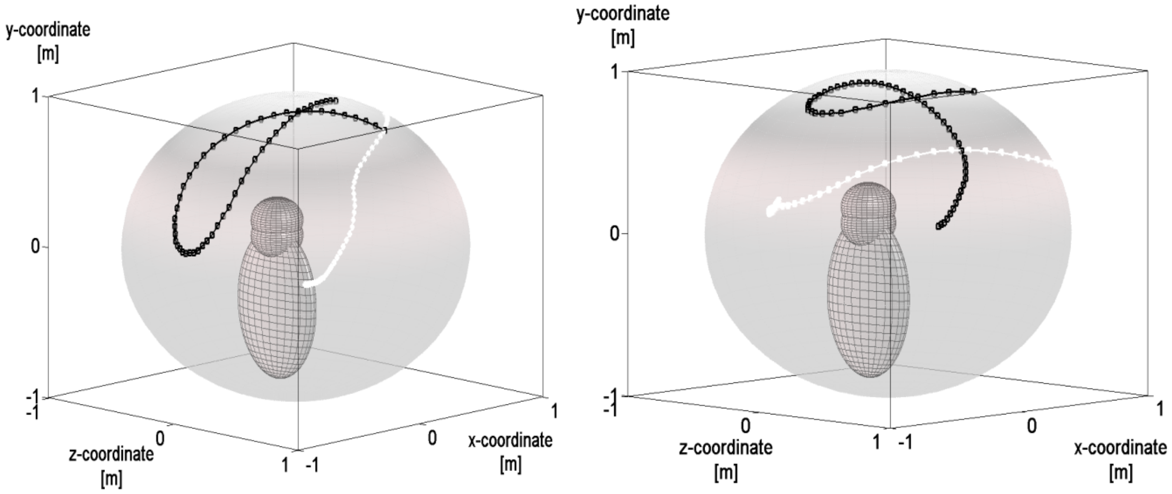

Figure 2: Trajectories of hand (black) and elbow (white) with respect to the shoulder joint, normalized and projected to a sphere

In conclusion: The advantage of applying the modelling method is to avoid the problem of gimbal lock while describing complex shoulder kinematics. This allowed for a detailed analysis of shoulder kinematics which revealed interdependencies of internal and external rotation with abduction and adduction movements during volleyball spikes

ANALYSIS OF LEG EXTENSION MOVEMENTS: The quantification of not directly-measurable attributes for muscle contractions in sport movements may lead to further information for strength training (Hatze 1998, Wank 2000). One area of interest is the comparability of internal versus external loads. Thus a neuromuscular multi body system model was developed for analyzing movements within the lower extremities (Roemer 2004).

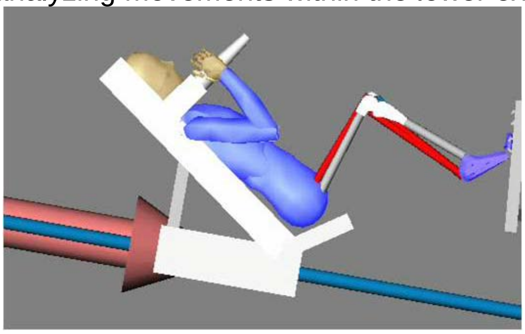

Figure 3: Simulation of leg extension movements in a leg press machine

While using inverse kinematics and inverse dynamics with a focus on the lower extremities, the knee model type is of major interest. To determine the accuracy of different knee joint model types and their influence on calculated muscle forces the model was evaluated as follows:

- accuracy of an magnetic resonance imaging (MRI) based knee joint model with respect to the coordinates of the moving joint axis.

- influence of individualized versus simplified model types on calculated muscle forces of a specific movement with respect to the patellofemoral joint.
- influence of individualized versus simplified model types on calculated muscle forces with respect to the knee joint.

The results indicated that it is necessary to use an individualized model that takes the moving joint axes of the knee joint and the patellofemoral joint respectively into account for the calculation of muscle forces for the Quadriceps. The consideration of the changing leverages within these joints during knee motion leads to more realistic results for calculated muscle forces. Otherwise it was not possible to calculate muscle stimulation functions that were comparable with measured EMG data (Roemer et al. 2010).

In order to simplify the process of gaining input data for the above described knee joint models, a new approach to estimate the moving joint axis within the knee joint using motion analysis data was evaluated. Therefore, a single case study was performed and data for the knee model were collected via MRI scans and motion analysis using a 12 camera VICON system (240Hz). MRI scans in 13 different knee positions were used to collect data representing the relative movement between tibia and femur for knee flexion movements.

Furthermore, the coordinates of 40 markers placed around the thigh and 40 markers placed on the shank were used to reconstruct the relative motion of these two segments. An optimization method described by Andriacchi et al. (1998) was used to minimize the influence of the skin movement on the marker coordinates. However, the results of this case study revealed that the approach to calculate the moving joint axis out of motion analysis data was insufficient. A combination of inaccuracies of the motion analysis data due to skin movements and numerical problems related to the kinematic theory approach caused this result. In conclusion: It is essential to use individually parameterized models for the knee joint as well as for the patellfemoral joint while analyzing the correlations between external and internal loads and the efficiency of specific training exercises for the lower extremities.

ANALYSIS OF ROWING MOVEMENTS (PILOT STUDY): Movement biomechanics of overweight and obese individuals are not well understood. One factor that has been linked to obesity and the disability to perform activities of daily living is osteoarthritis as well as other musculoskeletal diseases. It is well known, that a reduction in body weight will decrease the risk and/or severity of such diseases (Messier, Gutekunst, Davis, & DeVita, 2005). Caloric control and moderate exercise are crucial for reducing body weight in obese individuals. In general, non weight bearing exercises are recommended because it is assumed that they are more beneficial than weight bearing exercises. However, so far the literature reveals no clear evidence with respect to this assumption and there is little research available investigating the influence of obesity on joint mechanics. For this study the non weight bearing exercise of rowing on a Concept 2 (Model E) rowing ergometer was chosen. Ten normal weight and ten overweight volunteers with no or minor previous rowing experience took part in this study. The subjects were asked to row at three different resistance levels for two minutes each with two minutes of rest in between. Anthropometric measurements were taken, such as body weight, body height, and body composition. Three dimensional coordinates of the whole body movement were measured using a Vicon Motion Analysis System with 6 cameras. It was hypothesized, that obese subjects will show different results in leg joint angles than normal weight subjects. This two tailed hypothesis was chosen, because no previous data existed indicating a relationship with obesity and joint kinematics.

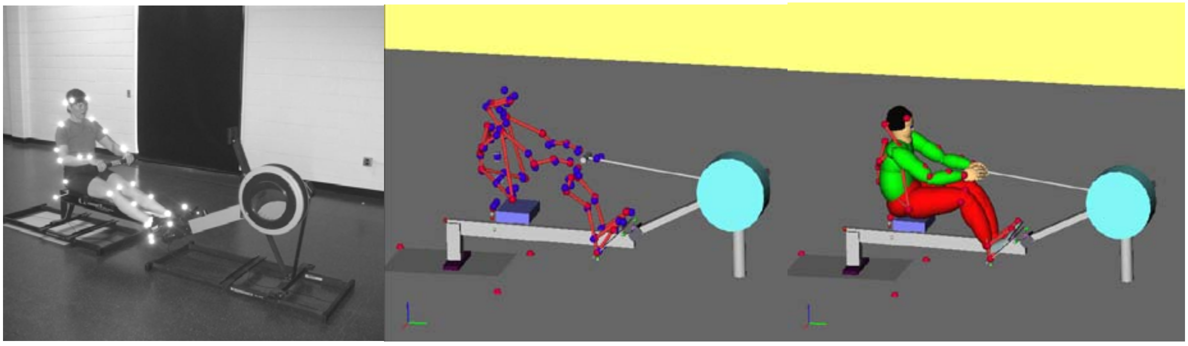

Figure 4: Process of data acquisition, transferring the movement data to the multi body system model, and reconstruction of the movement performing inverse kinematics.

The results of the flexion angles in hip and knee joints indicate that there is a relationship of body weight and movement kinematics. Both, the hip and the knee joints show differences with respect to maximum flexion angle. Obese subjects tend to have less hip and knee flexion in the catch phase. A follow-up study with 40 subjects was performed and first results will be presented by Richter et al. 2010.

CONCLUSION: The presented studies demonstrated how modelling methods can be applied to human movement analysis. It was shown how different joint models may influence results of calculated muscle forces and how different modelling methods can help describing complex shoulder movements.

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ACCOMMODATING STRATEGIES FOR PREVENTING CHRONIC LOWER EXTREMITY INJURIES

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Trauma or tissue damage is simply the result of applying too much stress to a tissue via an external load either directly or indirectly. The application of force to the system results in stress that can cause tissue damage, *i.e.*, injury. The problem can be viewed from two perspectives as stated by Nahum and Melvin in the preface of their book Accidental Injury, Biomechanics and Prevention (1993). One perspective is that of the professionals involved in injury diagnosis and treatment while the other is that of engineers and scientists (especially biomechanists) interested in the mechanics of injury. Both perspectives are well documented/represented in the professional and scientific literature.

Tissue damage/injury results when a tissue is stressed beyond some critical value/tolerance level. The stress is a result of the magnitude of the force, the type or direction of the force and the time interval between repetitive loading. Tissue damage can be acute (a single traumatic event) or chronic (developing/progressing over time). Most lower extremity injuries resulting from running and landing activities are chronic, resulting from repetitive loading to underdeveloped and/or unprepared structures. Chronic injuries can be avoided or minimized by adhering to a number of simple principles that are discussed in this paper.

Running is and has been one of the most popular forms of exercise and competition since the 1970s. During that time running mechanics has been a primary research focus in numerous biomechanics laboratories. Despite the extensive "scientific" literature that has been generated, relatively little, if any, progress has been observed when examining running injury statistics. Some representative injury statistics across the time span are given in Table 1.

Another statistic that has not changed appreciably relates to the most common running injuries. In a recent document from the American Orthopedic Society for Sports Medicine (AOSSM, 2008) the six most common chronic injuries are cited (Table 2). These are the same six common injuries identified by James et al. (1978) three decades earlier. A final comparison worth noting relates to the causes of injury. General agreement exists among publications by AOSSM (2008), Hoeberigs et al. (1982) and James et al. (1978) that the primary cause of injuries is and has always been training errors with anatomical factors and shoes and surfaces making a significant contribution.

Table 1

INJURY STATISTICS: RUNNING

KOPLAN et al, 1982	35%
LYSHOLM et al, 1987	65%
MARTI et al, 1988	46%
WALTER et al, 1989	48%
van MECHELEN, 1992	50%
ASPLUND, TANNER, 2004	50%

Table 2

Ten Common Injuries
(AOSSM, 2008; Others)

Plantar Fasclitis	Foot
Stress Fractures (Foot & Leg)	Foot / Leg
Achilles Tendonitis	Leg
Shin Splints	Leg
Iliotibial Band Syndrome	Knee
Patellofemoral Pain Syndrome	Knee
Ankle Sprains	Other
Muscle Pulls	Other
Blisters	Other

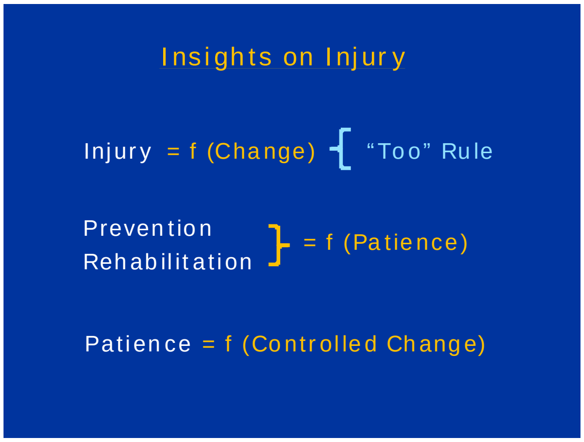
Why have we made so little apparent progress in our attempts to understand and prevent running injuries? Although most researchers claim to be evaluating similar/homogeneous groups of subjects this assumption must be questioned. Colonel John Stapp (Miller, 1979) observed that a human is a "fifty liter rawhide bag of gas, juices, jellies, gristle and threads movably suspended on more than 200 bones presiding over by a cranium, seldom predictable and worst of all living and presents a challenge to discourage a computer into incoherence." Given this expressed complexity do we really think that we can capture human performance in a simple model? Remember the human system feeds back on itself through continual experiences and changing perceptions never remaining in the same state. Is there such a thing as an average person/runner? If not, then it is unlikely that we can derive propositions about individuals from mean propositions derived from groups of people. Several examples will be discussed in the presentation.

Perhaps we have actually made more progress than the cited data suggest. During the past several decades many research studies have contributed to the scientific body of knowledge related to running mechanics and injury mechanisms. This knowledge certainly has positively influenced those professionals committed to the art and skill of diagnosing and treating running injuries. If this generated knowledge is being used by professionals to solve runners' problems more effectively and efficiently then science has made a positive contribution. Although no scientific data support this observation, anecdotal data suggest this to be the case.

CAUSE OF INJURY: The relationship between the prevention, cause and rehabilitation of injury is in the general sense relatively simple as shown in Figures 1 and 2. All are simply a function of change, however that change must be carefully controlled, *i.e.* avoiding too much, too soon, too fast, too similar, too different. The complexity of the problem results from the fact that we are all unique individuals structurally and functionally and have different experiences and goals that must be integrated into our exercise/running programs in an optimal satisfying way. In order to understand the complexity of this problem, it is important to examine selected aspects of change and their relationship to chronic injury.

Figure 1

Figure 2



PHYSIOLOGICAL ADAPTATION / ACCOMMODATION: The anatomical structures making up our system are in a continuing state of modification or adjustment as a result of environmental demands/stimuli.When these structures are modified positively the result is accommodation. Positive accommodation is the result of applying sufficient loads to obtain a physiological response for a specific activity (increase the acute threshold). The load must be varied enough to avoid cumulative stress/injury (increase chronic threshold). If the stress exceeds the tissue's physiological ability to accommodate, the tissue will be modified negatively becoming damaged resulting in injury. Remember, training errors account for as much as 60% of running injuries as a result of stressing the system with forces that are beyond its current physiological threshold, *i.e.* too much, too fast, too soon, too similar, too different. The physiological response of a tissue is constantly changing and at any instant in time is dependent upon such factors as age, gender, rate of loading, loading duration, and history (experience, tissue use or disuse).

Some response patterns are shown in Figure 3. The goal of any training program should be to increase the tissue threshold over time in addition to elevating the threshold overall. When the tissue is stressed beyond its ability to maintain or increase its threshold, the curve begins to plummet rapidly indicating the onset of tissue failure/injury. Preventing this rapid decrease is unlikely without rest and a significant reduction of the stressor.

MOVEMENT / MOVEMENT STRATEGIES: From a behavioral perspective movement is simply a tool for problem solving. The primary constraint of any movement pattern is the task or goal which is further constrained by our morphology, the current environment and the applicable biomechanical principles.

We are all creatures of experience turning into memories which determine our perceptions and expectations which in turn dictate our movement assumptions and actions. Our myriad of experiences combined with the complexity of our neuro-musculo-skeletal system suggests that we have multiple movement patterns to choose from to accomplish any particular task. We are able to select a particular movement strategy (a neuro-musculo-skeletal solution) within certain constraints to perform the motor task. As an example, Figure 4 shows possible response patterns to the impact forces during running or landing as a result of added load to the extremity. The overall dimensions of the parallelogram are determined by strength and tissue threshold. A pure Newtonian response results when the added load is ignored (the diagonal line). A pure neuro-muscular response results when no increase in impact force is observed as a result of an accommodating motor pattern (the horizontal line). All responses within the parallelogram are the result of a partial acknowledgment of the added load producing a complex biomechanical response. Additional response patterns can become available as shown in the figure to the right by increasing strength and/or tissue threshold.

Figure 3

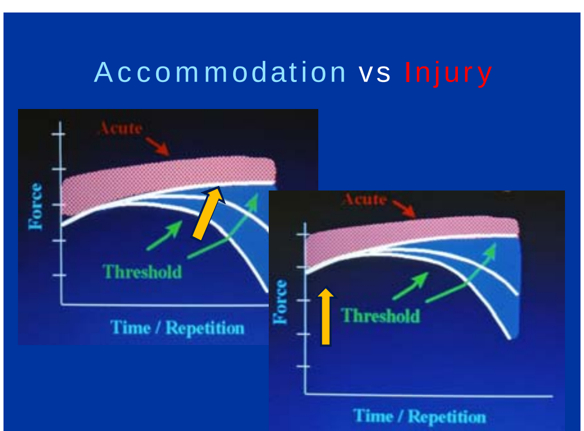
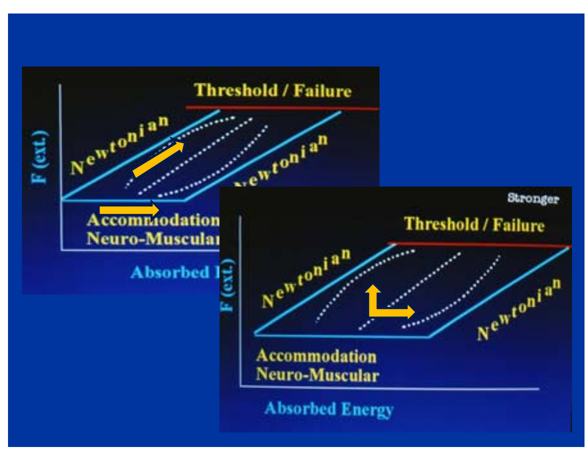


Figure 4



VARIABILITY*:* As previously stated injuries are caused by changes in force/stress. Injury prevention and rehabilitation are also the result of changes in force. The magnitude and/or direction of forces can be changed by variations within activity. Variations within activity can be accomplished by changing factors in the environment such as shoes, surface and terrain. A second option is to change performance characteristics such as foot strike pattern, knee angle, etc. Variations can also be accomplished by changing the activity, *i.e.* running, cycling, cross training. The purpose of these variations is to broaden the normal and healthy region of performance as shown in Figure 5 since the narrower the band of performance the more susceptible one is to chronic injury. Expanding the band of performance, however, must be accomplished very gradually since rapid change is also a risk factor. Another area of concern regarding this relationship is how one trains.

SPECIFICITY VERSUS VARIABILITY: Performer variability is inherent in all human movement as a result of system complexity and random perturbations. From an injury perspective this is a positive feature since it helps minimize chronic injury potential. For some activities such as gymnastics where the goal is precision/replication variability has negative connotations. In running, the elimination/minimization of variability is probably more critical for sprinters than distance runners. Relative to training and elite performance a certain amount of specificity is essential but specificity also increases the risk of chronic injury. Figure 6 shows the relationship between performance variability and risk of injury (left) and performance variability and skill development (right). What this figure illustrates is that risk reduction decreases dramatically with only small increases in variability while decreases in skill development are far less dramatic for similar increases in variability. The important point to be made is that it is important to consider this relationship when assessing performance goals. Remember, an injured athlete never improves performance.

Figure 5

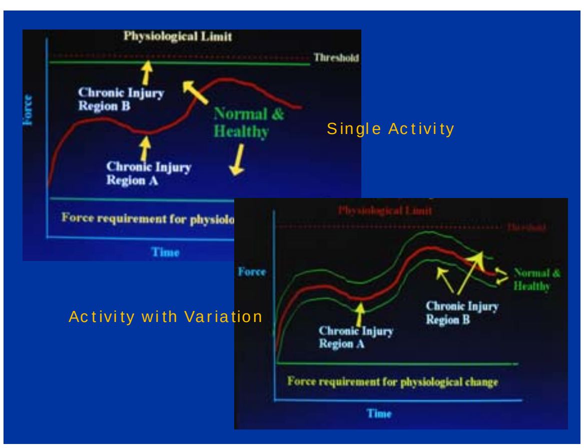
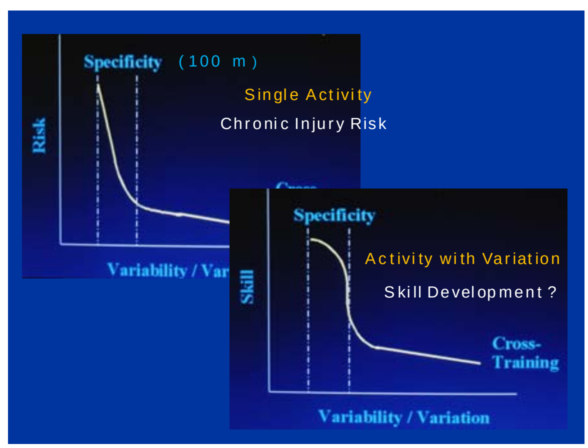


Figure 6



SUMMARY: Take-home lesson 1 is to acknowledge the possible dilemma between skill performance and injury risk. Each individual must weigh the choices among injury risk, performance and goals. Take-home lesson 2 is to remember that injury is a function of change, *i.e.* too much, too soon, too fast, too similar, too different. Just as injury is a function of change so are injury prevention and rehabilitation. All change must be carefully controlled through patience. Although this discussion tended to focus on running, it is equally applicable to other types of repetitive training and exercise. Some final suggestions include (1) set appropriate goals, (2) train/exercise smart, (3) get adequate rest, (4) do not ignore pain, (5) always think prevention, and (6) have fun.

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PARALYMPIC SPORTS, THE NEXT FRONTIER FOR SPORTS SCIENCE

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The Paralympic Games is the pinnacle of sport for many athletes with a disability. The purpose of this paper is to briefly provide some background on the Summer Paralympic Games and their eligibility and classification rules. Results from selected studies examining the biomechanics of locomotion (amputee running, swimming and wheelchair pushing) and projecting external objects (e.g. throwing and hitting) as well as the evolution of sports performance and training practices such as strength and conditioning will be described. Recommendations for how this evidence can be used to improve athletic performance in Paralympic sports and inform future research are also provided.

KEY WORDS: adapted physical activity, biomechanics, sport for the disabled.

INTRODUCTION: Sport has not been traditionally advocated or emphasized for people with disabilities. This began to change in the middle of the 20th century with organised sport for these individuals beginning in Stoke Mandeville, England in 1948 (Bailey, 2008). This lead to the development of various organisations including the International Stoke Mandeville Games Federation in 1959, International Organization of Sport for the Disabled in 1964 and the International Paralympic Committee in 1989 (Bailey, 2008). As a result, Paralympic Games are now held directly after the Olympics Games, with the Paralympics held at the same host city as the Olympics from 1988 (International Paralympic Committee, 2009). The growth of the Paralympics Games is demonstrated by the fact that in the 2008 Beijing Paralympics there were 3,951 athletes from 146 countries competing in 20 sports (International Paralympic Committee, 2009). With the growth of the Paralympic Games, so too has society's views on these athletes changed. Instead of being viewed as people "suffering" from a disability, they are now often seen as inspiring high performance athletes. Such changes in public perceptions may have also altered the research emphasis, whereby many more research studies are now being conducted with a sports performance rather than rehabilitative focus. The remainder of this paper will focus on Parlympic athlete eligibility and classification, the biomechanics of Paralympic locomotion and projecting external objects, evolution of Paralympic sports performance as well as contemprary training practices.

ATHLETE ELIGIBILITY AND CLASSIFICATION: A wide variety of athletes can compete in Paralympic Games. In the 2008 Beijing Paralympic Games, classifications groups included spinal injury, amputee, visually impaired, cerebral palsy and les autres (International Paralympic Committee, 2009). As a result of the within- and between class heterogeneity of athletes, ongoing debate exists on how they should best be classified so to ensure fair competition (Jones and Wilson, 2009; Burkett, 2010; Tweedy and Vanlandewijck, in press).

LOCOMOTION: Paralympic athletes compete in a variety of sports where they need to move quickly when running, swimming, cycling (leg- or arm-propelled) or pushing a wheelchair. The following section will examine the findings of some selected studies that examined the biomechanics of amputee running, jumping, swimming and wheelchair propulsion.

Running and Jumping: The biomechanics of amputee running have recently been debated with allegations that individuals like the double trans-tibial amputee sprinter Oscar Pistorius who was attempting to qualify for the Beijing Olympic as well as Paralympic Games in 2008 was at an advantage over his intact-leg rivals due to his prosthetic limbs (Nolan, 2008; Burkett, 2010). As a result, Weyand et al. (2009) compared the biomechanical and physiological demands of Oscar Pistorius to elite intact-limb male 400 m sprinters. It was found that Oscar Pistorius had a similar or slightly lower metabolic cost during sub-maximal running, similar sprinting endurance but considerably different running mechanics than the intact-limb sprinters. Such results appear similar to the conclusions of Nolan (2008) in which after reviewing the literature, it was found that while there are considerable differences in the way that amputee and intact-limb sprinters actually sprint, there is little evidence to support the view that amputee runners are at any form of advantage. Several studies have also examined amputee jumping. During a study of the biomechanics of amputee long jump, Nolan and Lees (2007) observed many significant kinematic differences (e.g. hip and knee angle, height of centre of mass, stride length etc) at touchdown and takeoff of the last three strides prior to jumping for trans-femoral and trans-tibial long jumpers, with these values often different to that of intact-limb jumpers of similar standard.

Swimming: Several studies have been conducted involving Paralympic swimmers. Daly and colleagues (2001; 2003) found that Paralympic swimmers exhibit very similar race patterns to Olympic 100 m swimmers and that clean swimming, turning and finishing speed were highly correlated with race results, although the key phase was the second half of each 50 m lap. Daly et al. (2003) also observed that between-swimmer differences in 100 m speed were more related to stroke length than stroke rate, whereas within-swimmer changes in race speed were more related to changes in stroke rate. These results were extended by Osborough et al. (2009) who investigated how selected anthropometric characteristics were associated with the stroke length, stroke rate and swimming speed of 13 unilateral arm amputee swimmers. Similar to previous studies, it was found that within-swimmer changes in maximal speed were more associated with stoke rate than length. While no significant anthropometric correlates were found for stroke length, stroke rate at maximal speed was found to be significantly correlated to biacromial breadth ($r = 0.86$), shoulder girth ($r = 0.64$) and upper arm length ($r = 0.58$) (Osborough *et al.*, 2009).

Wheelchair: Wheelchairs are used for Paralympic athletic race events e.g. 100 m up to marathon and team sports such as wheelchair basketball, rugby and tennis. Some excellent, although slightly dated reviews of wheelchair locomotion are available (e.g. van der Woude *et al.*, 2001; Vanlandewijck *et al.*, 2001). One of the key findings of these reviews was that as the velocity of wheelchair locomotion increases, the propulsive style changes from circular to pumping, push time is significantly reduced, while the recovery time and push angle are relatively unchanged. These results have implications to training programs for these athletes as well as to the applicability of general research in this area to high performance sport.

Wheelchair sporting performance is directly influenced by the combination of three factors, the athlete, wheelchair and the wheelchair-athlete interface (Vanlandewijck *et al.*, 2001). These three factors in turn can directly influence the kinematic, kinetic and electromyographic characteristics of wheelchair propulsion, the degree of friction and air resistance encountered and the ability of the athlete to perform sport-specific activities e.g. catching, passing and hitting. As a result, wheelchair design continues to evolve, with a variety of specific chairs used across and even within Paralympic sports (Burkett, 2010). These specific chairs are required as unlike wheelchair race events, wheelchair team sports are characterized by frequent acceleration, deceleration and change of direction as well as requiring the athlete to hold a racquet and/or catch, pass or hit balls (Goosey-Tolfrey and Moss, 2005; Reid *et al.*, 2007). Unfortunately, little research has been conducted on the biomechanics of wheelchair team sports. Goosey-Tolfrey and Moss (2005) examined the effect of holding a tennis racquet on 20 m wheelchair sprint performance from a stationary position. The tennis players were significantly slower when holding their tennis racquet, with this effect most noticeable over the first three strokes (Goosey-Tolfrey and Moss, 2005). Such results may have implications for the design of on-court conditioning sessions for these athletes as well as for the tactics used during competition i.e. how to hold the racquet during the first few strokes when attempting to reach a ball hit by their opponent.

PROJECTING EXTERNAL OBJECTS: Even though many Paralympic sports involve projecting external objects e.g. throws and/or hits in athletics and team sports such as

wheelchair basketball, rugby and tennis, there appears to be much less research in this area than that for locomotion. Chow et al. (2000) examined the 3-D kinematics of 17 wheelchair shot putters and found that the average speed and angles of release of the shot were smaller than those of Olympic throwers. Significant correlations were observed between the height of release, angular speed of upper arm at release, shoulder girdle range of motion during delivery and average angular speeds of the trunk, shoulder girdle and upper arm during delivery to the classification and distance thrown (r = 0.52-0.79). Relatively similar results have also been observed for wheelchair javelin performance by Chow et al. (2003), suggesting that release parameters and the ability of selected upper body segments to obtain high angular speeds are critical for success in wheelchair shot put and javelin.
When performing flat and kick serves, Paralympic wheelchair tennis players produced significantly less racquet velocity and utilized somewhat different shoulder and trunk ranges of motion than their able-bodied peers (Reid *et al.*, 2007). However, as the Paralympic and able-bodied groups experienced relatively similar shoulder joint kinetics, both groups of tennis players may be at similar risks of shoulder injury when serving (Reid *et al.*, 2007). When comparing novice and elite wheelchair tennis players preparing to return serve from either a video or real opponent, Reina et al. (2007) found many significant differences. For example, during the ball toss the elite players focused more on the free arm and head/shoulders of the server. The elite players also had quicker reaction and movement times when facing serves from the real, but not video opponent. These results suggest that elite players are able to generate faster reaction and movement times, and ultimately likely improve their return of serve by utilizing some visual cues from their opponent.

EVOLUTION OF SPORTS PERFORMANCE: As the opportunities for people with disabilities to compete in sport at an elite level continues to increase, so too does their level of performance, with a total of 279 world records set at the 2008 Beijing Paralympics Games (International Paralympic Committee, 2009). However, only one scientific study has investigated the progression and variability of Paralympic sports performance. Using a sample of 120 male and 122 female Paralympic swimmers, Fulton et al. (2009) calculated the annual progression and variability in 100 m freestyle performances over a three year period. As between-competition variability and annual progression in performance were ~1.3% and ~0.5% respectively, Fulton et al. (2009) stated that Paralympic 100 m swimmers would need to improve by at least 1-2% annually to increase their medal chances.

TRAINING PRACTICES: Little research has examined training practices or the effects of strength and conditioning on Paralympic sport performance. Fulton et al. (2010) quantified the training of 16 Paralympic swimmers in the final 16 weeks of training prior to a World Championships. While these swimmers performed less weekly training volume than that of Olympic swimmers, they followed a similar periodized plan with respect to changes in volume and intensity. Fulton et al. (2010) however speculated that the Paralympic swimmers could have benefited from a more substantial taper prior to competition. Turbanski and Schmidtbleicher (2010) compared the effect of eight weeks of moderate load bench press training on the upper body performance of wheelchair athletes and college students. While both groups significantly improved bench press peak force, 1RM strength and maximum rate of force development, there were some trends ($p < 0.10$) for these effects to be greater in the wheelchair athletes and for the wheelchair athletes to improve 10 m sprinting speed.

CONCLUSION: The number of relevant articles found when completing this mini-review suggests that the majority of the research into the biomechanics and physiology of adapted physical activity has concentrated on non-athletic individuals performing activities of daily living rather than Paralympic athletes performing their sports-specific skills. Results of the limited sport performance studies suggest that while there are many similarities in the biomechanics and physiology of Paralympic and Olympic athletes, there are also many significant differences. Coaches and sport scientists who work with Paralympic athletes will therefore need to be aware of these similarities and differences if they are to contribute to the continual development of their athletes and to the overall evolution of Paralympic sports performance. Finally, I hope that this presentation has highlighted the opportunities for sport scientists to work with Paralympic athletes in either a sport science support or research role.

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BIOMECHANICAL FACTORS IN SPRINT TRAINING- WHERE SCIENCE MEETS COACHING.

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This paper examines the biomechanics of sprinting and sprint training. Various biomechanical models of sprint performance are considered with respect to the start, acceleration and speed maintenance phases of the 100 m sprint event together with the research that underpins those models. The impact of research on strength and conditioning training is discussed with special reference to the control of leg-spring stiffness and the applications of resistance and complex training modalities. Training practises for sprinting are discussed with respect to scientific evidence. The relevance of commonly used sprint and running drills is evaluated in relation to the kinematics and muscle activation patterns in sprinting. Finally, a simple coaching related model for the development of sprinting is presented which is consistent with scientific evidence.

KEY WORDS: Running, leg spring stiffness, complex training, reactive strength

INTRODUCTION: The mission of ISBS emphasises the importance of ‘bridging the gap between scientists and practitioners’. Sprint running is fundamental to successful performance of many sports activities and athletics events. Consequently, the biomechanics of sprinting has been researched in great detail over many years. Despite the considerable body of scientific knowledge currently available on the biomechanics of sprinting and sprint training, Jones et al, (2009) found a dearth of knowledge amongst expert coaches on the technical constructs which govern the successful completion of each phase of the 100m sprint event. It is therefore apparent that much more work is required to bridge the gap between scientists, sprint coaches and their athletes. It is intuitively obvious that the quality of performance in sprinting or any other activity must be related to the quality of the training experience and preparation of the athlete. Therefore, consideration must be paid to the biomechanical factors that improve sprint training as well as those that affect sprint performance. This paper summarises the evolution of biomechanical “technical knowledge” of sprinting and sprint training. It examines the biomechanical and technical models of sprinting and considers biomechanics factors in sprint training. These factors focus on how research in biomechanics has underpinned improvements in strength and conditioning training as well as improvements in running technique. Finally, a simple technical model of sprinting is presented which is underpinned by scientific evidence from biomechanical investigations and reviews. The paper and accompanying presentation examines in detail, how practises should be structured and developed to achieve optimal sprinting technique.

BIOMECHANICAL MODELS OF SPRINTING: Biomechanical models of sprinting have been presented in various forms and each model contributes to our understanding of the biomechanics and performance of the activity. The Deterministic Models of Hay and Reid, (1988) show the factors that affect performance of sprinting and the relationships between those factors using simple mathematical relationships. The advantage of this approach lies in the ability of the models to identify the biomechanical factors that truly limit performance. The disadvantage is that many of the factors may not be readily observable in field situations and therefore alternative coaching models are presented in the form of picture sequences and descriptions of critical features, (see figure 1). Graham and Harrison, (2006) have provided a typical deterministic model for the start and first 5 m of the sprint event, (see figure 2). While this model was derived to identify factors limiting performance of the start to 5 m, the right side of the flowchart identifies the deterministic factors that act throughout the race after block clearance.
Maximum Velocity Models: When the athlete has accelerated to maximum running velocity, (which is attained after 25 to 50 m depending on performance level), then performance is limited by the ability of the athlete to maintain speed.

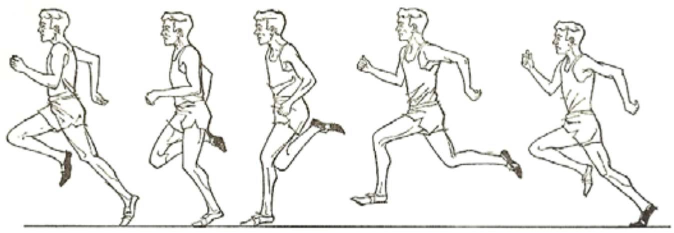

Figure 1: Picture sequence for sprint running. (From Dyson, 1973)

Research has shown that in maximum speed running, the leg action can be effectively modelled as a simple linear spring-mass model, (Farley et al, 1991; Farley et al, 1996; McMahon et al, 1987). The stiffness of the leg spring, K_{leg} can be determined by dividing the peak ground reaction force by the change in leg length, ΔL and the vertical stiffness, K_{vert} equals the peak vertical ground reaction force divided by the vertical displacement of the centre of mass, Δy. In vertical hopping $K_{vert}=K_{leg}$. Over the last 20 years, various researchers have demonstrated the importance of leg-spring and vertical stiffness in fast running. Arampatzis et al, (1999) found that leg stiffness increased with increasing running speed and Farley et al, (1991) found that the stiffness of the leg-spring can change as much as twofold to accommodate different hopping frequencies.

Chelly and Denis, (2001) showed that leg stiffness values are correlated to maximal running velocity (r = 0.68, $P < 0.05$). Similarly, experimental research on well trained athletes has also shown that sprinters have significantly higher leg spring stiffness compared with distance runners (Harrison et al., 2004). It is therefore clear that the ability to interact with the ground using a stiff spring-like action of the lower limb, is critical in encouraging a fast cadence and running speed in the speed maintenance phase of sprinting.

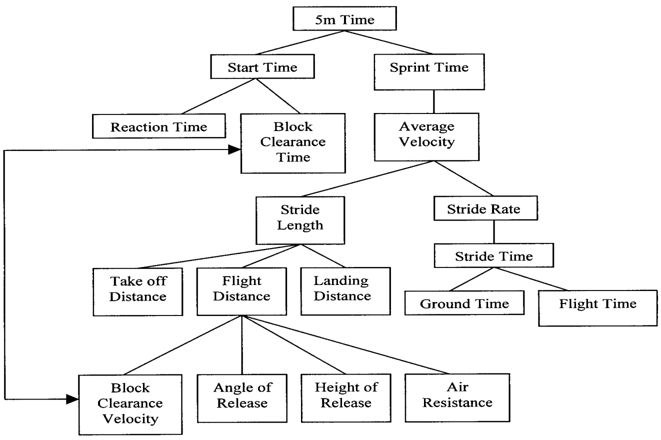

Figure 2: Deterministic model for sprinting based on the guidelines of Hay and Reid (1988).

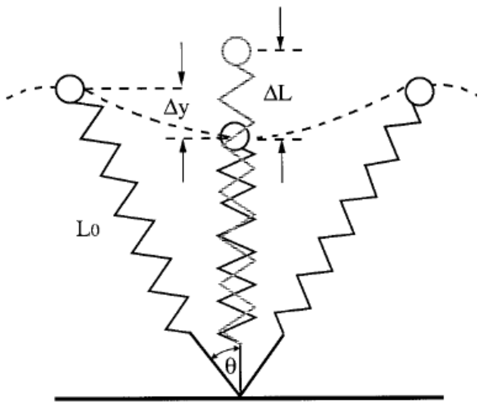

Figure 3: Diagram of the spring-mass model during running L_0, ΔL, Δy, and θ represent the initial length of the leg spring, the maximum compression of the initial length of the leg spring, the maximum vertical displacement of the centre of mass, and the half of the angle swept by the leg spring during the stance phase, respectively. The spring–mass model drawn with a discontinuous line represents the uncompressed leg spring. The arc drawn with a discontinuous line represents the path of the mass (centre of gravity of the subject) during the stance phase. (From, Chelly and Denis, 2001)

BIOMECHANICAL FACTORS IN SPRINT TRAINING: Research in biomechanics points to the importance of strength development in the early acceleration phase and the development reactive strength or leg stiffness for maximum speed running. The use of sledge towing and resistance running has been found to benefit the early acceleration phase of sprinting. Cronin et al, (2008) established that sledge towing was an appropriate training modality for the early acceleration phase in sprinting, provided the amount of resistance was moderate and did not result in major changes in the joint kinematics of the sprinting action. Harrison and Bourke, (2009) found that sledge training with a load of approximately 13% of body mass, significantly improved 5 m and 10 m movement times in field tests and improved starting strength in laboratory-based tests. It appears then, that resistance running exercises are a valuable training modality for developing the early to mid-acceleration phase. The transition between early acceleration and the attainment of maximum velocity, which is commonly referred to as the ‘pick-up’ or ‘drive phase’ is an important aspect of sprinting but to date there is little scientific work on the biomechanics of this phase. However, logic points to the use training exercises where resistance is applied during running and then released and this is borne out by observation of coaching practices.

In recent years, several studies have examined the application of complex training (which alternates heavy resistance and plyometric exercises) to improve performance in sprinting and jumping activities. The outcome of complex training studies have been somewhat conflicting with some authors finding clear benefits (Robbins, 2005; Comyns et al, 2006; Chatzopoulos, 2007) and others observing no perceptible benefits (Ebben and Watts, 1998; Jones and Lees, 2003; Scott and Docherty, 2004). Close inspection of the research in complex training reveals wide variations between studies on how data are analysed and this may account for the inconsistency in findings. Comyns et al, (2006 and 2007) showed that complex training could be used to induce acute changes in the leg-spring stiffness response during rebound jumps. Comyns et al (2007) indicated that the optimal load for the heavy resistance component of complex training was >90% of a 1 repetition maximum. The optimal recovery interval for inducing a post activation potentiation response appeared to be highly individualised (Comyns et al 2006). Comyns et al (2010) found that the benefits of using complex training to improve sprinting performance were highly variable and that athletes probably required sustained and repeated exposure to this training modality to gain significant benefits. In summary, it appears that a major benefit of complex training is in improving the leg-spring stiffness response and reducing contact times during running and jumping.

USE OF ISOLATION DRILLS: An important feature of sprint coaching is the establishment of optimal movement and coordination patterns. The predominant coaching model for this derives mainly from descriptive movement sequences and critical features. Coaches and athletes often use a variety of running drills (sometimes called isolation drills) to encourage the development of optimal movement and coordination patterns. These isolation drills are designed to help the athlete to practise specific parts of the running skill and it is therefore assumed that the drills are the parts of a whole-part-whole learning strategy. For this approach to be successful, it is important that the part practises relate well to the correct sprinting techniques and activate the muscles in patterns that are consistent with sprinting. From a pedagogical perspective, the use of varied part practices is well justified provided the movement parts relate well to the whole skill. The use of such drills is widespread but close inspection of some of the drills reveals they have questionable relevance to sprinting. A typical example of inappropriate practise is the use of the heel flick drill which is assumed to mimic the knee flexion action during the early swing phase of sprinting.

Figure 1 shows a typical picture sequence for sprinting and it is clear that soon after toe off the knee and hip joints appear to flex as the leg moves through the swing (or recovery) phase. Dyson, (1973) even implied that this knee flexion occurred to decrease the load on the hip flexors and increase the speed of the swing phase.

“In this way the mass of the leg is brought closer to the hip axis, reducing

the leg's moment of inertia and increasing angular velocity." (Dyson, 1973)

Harrison and Warden, (2003) have pointed out that this exercise may be of dubious benefit in sprint training because careful observation of the sprinting shows the heel flick action is not consistent with the pattern of movement of sprinting. In the swing phase of sprinting, the knee does not flex before the thigh is flexed. In reality the knee flexion occurs simultaneously with hip flexion or slightly after the initiation of hip flexion. Furthermore, various authors have shown that the hamstrings do NOT actively contract to facilitate knee flexion immediately after toe off. Weimann and Tidow (1995), Thelen et al (2005a) and Thelen et al (2005b) provided EMG data that demonstrated the hamstring muscles remain relatively inactive during the early part of the swing phase, (see figure 5).

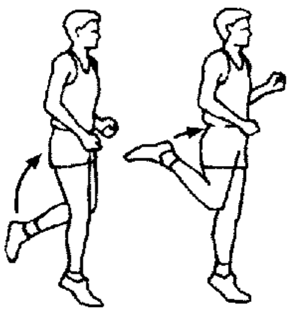

Figure 4: The heel flick drill commonly used in sprint training. (From Carr, 1999).

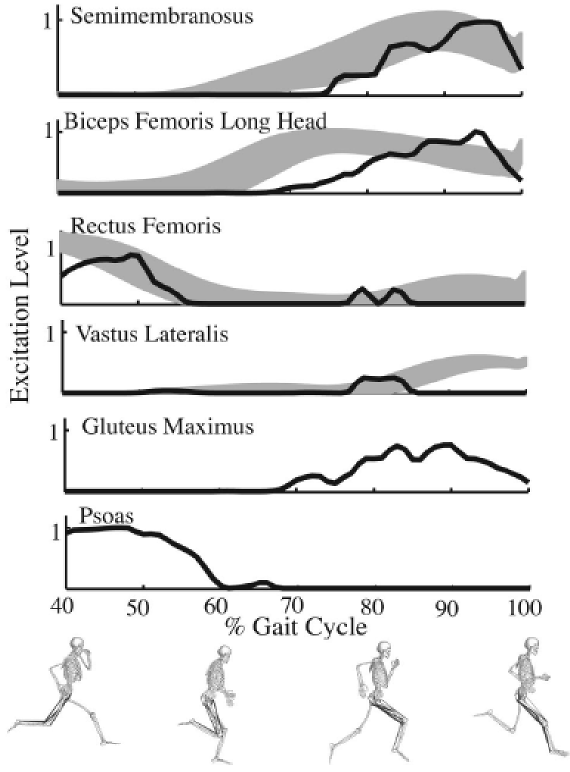

Figure 5: Simulation of a 9.3 m.s^{-1} sprint running cycle showing that the model predicted hip and knee angle kinematics close to experimental values throughout swing phase. The predicted excitations of the Semimembranosus m. Biceps femoris m. Vastus lateralis m. and Rectus femoris m. are compared with measured EMG activities (*shaded curves* are the mean ±1 SD of the rectified EMG activities for five subjects). The simulated activation sequences show that the hamstrings and gluteus medius m. are relatively inactive during the early swing phase. The Vastus lateralis m. and Psoas m. (hip flexors) are active in early swing phase. Note that the simulations also provide excitation estimates for muscles (e.g., psoas m.) that cannot be monitored using surface electrodes. (From Thelen et al, 2005b).

Johnson and Buckley (2001) examined hip and knee joint moments and powers during the mid-acceleration phase of sprinting and found that the knee flexion moments were relatively small just after toe-off. Considered together, these studies indicate that the hamstrings are not likely to be active immediately following toe off and therefore the practice of heel flick drills is inappropriate. Thelen et al (2005b) also suggest that hamstring activity at this point in the running gait cycle could increase the risk of injury.

Science contributes significantly to our understanding of the technical aspects of sprint training and performance but ultimately coaches need to distil the findings of scientific research down to simple technical models that emphasise the correct or optimal movement, coordination and muscle activation patterns for sprint performance. Knowledge of such optimal patterns may allow the coach to select systematically appropriate drills and training practises. Based on many years of experience of coaching international level sprinters, the author has developed a simple model of sprinting which has important implications for training. This model assumes that the arms play a subordinate, counterbalancing role to the leg action in sprinting and therefore, little emphasis is placed on correction of arm actions unless they are demonstrably unbalancing or destabilising the overall movement of the sprinter. This is somewhat contrary to the views of many expert coaches but is consistent with the scientific literature (Jones et al 2009). The primary emphasis in sprinting should be on the action of the legs which involves moving each leg alternately from the positions of the right leg in figure 6A and 6B. The emphasis during ground contact should be on the production of a short, pawing movement with stiff spring-like rebounding action during the speed maintenance phase. The relevance or otherwise of various drills and practises will be explored during the presentation using video sequences.

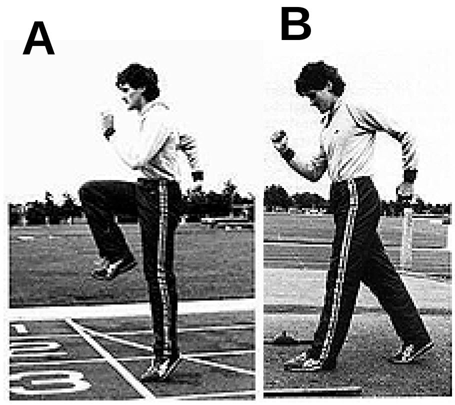

Figure 6: Two key positions in sprinting.

CONCLUSION: Effective coaching requires that coaches have a clear and valid model of sprinting technique that describes desired movement, coordination and muscle actions which are consistent with evidence from scientific investigations. It is concluded that in many instances, current coaching practise does not draw effectively on research findings on sprinting biomechanics. A simple technical model of sprinting is presented together with a series of relevant practises and drills.

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KICKING IN SOCCER

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Kicking as the defining action of soccer has gained huge interest within the scientific community with respect to practice, coaching, science and technology. This paper focuses on current knowledge of soccer kicking, including its biomechanical description, skill execution, kicking techniques, performance criteria, and measurement technology. Additionally, it refers to a fairly new research aspect, the influence of soccer footwear on the kicking movement and consequently on kicking success as it has been shown that soccer footwear alters the biomechanics of kicking. Finally, future directions of research are suggested that are helpful to enhance the fundamental understanding of kicking.

KEY WORDS: kicking technique, velocity, accuracy, measurement technology, footwear

INTRODUCTION

Soccer kicking is described as a complex motor movement consisting of six important stages: Approach angle, plant foot forces, swing limb loading, flexion at the hip and extension at the knee, foot contact with the ball, and follow-through (Barfield, 1998). The kicking procedure may also be divided into five essential aspects: Approach, support leg, kicking leg, foot to ball interaction, and ball flight (Lees et al., 2010). The resultant ball movement of soccer kicking is finally determined by its core phase characteristics, the foot to ball interaction, which only lasts about 10 ms (various authors). Thereby, the impact phase of soccer kicking is characterized as a mixture of impact-like and throwing-like mechanisms (Tsaousidis & Zatsiorsky, 1996). Shinkai et al. (2009) elaborated on these considerations and identified the four following specific phases of foot to ball interaction: phase I – centre of ball gravity (CBG) moves without ball movement (pure ball deformation), phase II – start of ball movement until ball velocity exceeds foot velocity, phase III – start of ball decompression with continuing decrease of foot velocity and further increase of CGB velocity, phase IV – foot loses ball contact while foot deceleration and ball acceleration stops (Figure 1). For powerful kicking Kellis & Katis (2007) have identified key aspects in the literature: Technique, optimum transfer of energy between segments, approach speed and angle, skill level, gender, age, limb dominance, maturity, the characteristics of foot to ball impact, muscle strength and power of the players, and type of kick. Accurate kicking has received less attention, important aspects are: Approach speed, kicking velocity reduction, foot to ball contact point, and ball spin (Kellis & Katis, 2007).

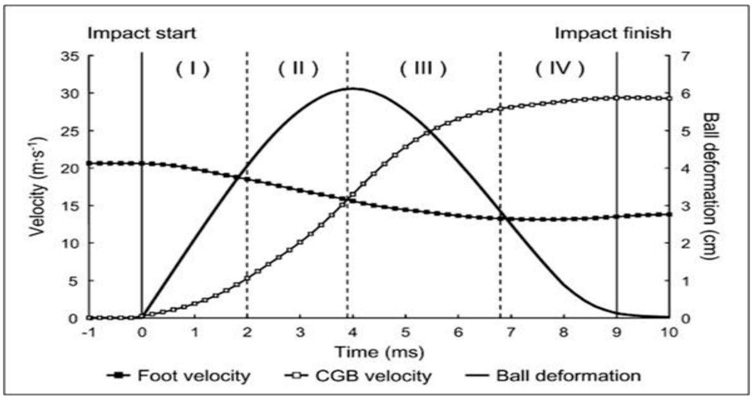

Figure 1: Foot-ball interaction (Shinkai et al., 2009)

KICKING TECHNIQUES

There is a selection of kicking techniques in order to cope with the demands of various specific game situations. The most obvious differentiation refers to kicking stationary or non-stationary balls. Thereby, non-stationary ball kicking situations are characterized by rolling or bouncing balls, the latter require volley or drop kick techniques. Player's kicking technique selection should be based on the specific necessity of ball speed, accuracy, spin, and ball flight and should be analyzed in relation to the player's position to the ball and on the field.

Kicking techniques may be divided into instep, side foot, and toe kicks. Thereby, the inner and outer instep kick aim to rotate the ball by the tradeoff of some velocity (Neilson & Jones, 2005).

PERFORMANCE CRITERIA

Important performance variables are ball velocity and accuracy, and also ball flight. A curved ball flight is achieved by imparting spin to the ball due to eccentric foot-ball contact. Maximum ball speed has been widely accepted to be the main biomechanical indicator of kicking success. However, there are good reasons to reconsider this notion as kicking accuracy might be much more important, as it applies to almost all passing and kicking throughout the game regardless of technique and power applied. When shooting on goal, players often want to maximize both, ball velocity and accuracy, to reduce reaction time of the goal keeper and to increase his distance to the ball. Obviously, this combination is difficult to achieve. It also occurs in various other sports like tennis, volleyball, or handball. Commonly it is referred to as the speed-accuracy-trade-off which has been quantified for the three most frequently used techniques in soccer (Figure 2; Sterzing et al., 2009).

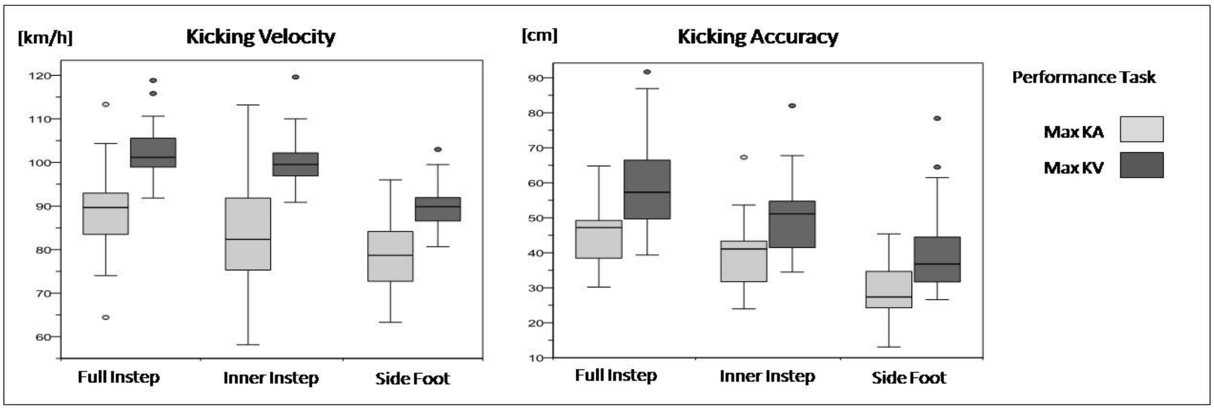

Figure 2: Max KA/ KV: Maximization of accuracy/ velocity (Sterzing et al., 2009)

Players kicked slower when trying to maximize kicking accuracy. Thereby, reduction of velocity was fairly similar across the three kicking techniques. Kicking accuracy was reduced when players kicked as fast as possible. Looking at kicking variability, players used a relatively stable velocity within each technique and performance task. In contrast, accuracy outcome appeared relatively variable. It was concluded that players can better tune their input velocity variable than controlling their accuracy output. Further performance criteria like ball flight and ball spin are more difficult to measure. For curved balls an important variable is the lateral deflection of its flight. This allows players to kick the ball around a defensive wall during a free kick more easily. For analysis of irregular and unsystematic ball flight curves, objective performance criteria are difficult to obtain.

MEASUREMENT TECHNOLOGY

Measurement technology is an essential aspect for the analysis of kicking. For testing kicking accuracy of players' motor performance in practice sessions, discrete target areas are commonly used. However, for obtaining more precise and metric data, advanced measurement technology is needed. For precise measurements of ball accuracy different measurement procedures have been introduced: Ball prints produced by carbon paper on a wooden wall (Finnoff et al., 2002), a custom made circular electronic target (Hennig et al., 2009), and high-speed video capturing (Sterzing et al., 2009). All these provide precise ball accuracy data with high resolution, however the latter two needing a certain amount of technical instrumentation. For kicking velocity measurements commonly high-speed video capturing or radar guns are used by various researchers. Ball spin can be measured by motion analysis of an instrumented ball with retroreflective markers attached. An extraordinary way of showing ball flight was presented by Asai et al. (2007) visualizing the knuckle effect of a ball that was kicked with no rotation by use of titanium tetrachloride.

SOCCER FOOTWEAR

Soccer footwear has to be regarded as an artificial and beneficial interface between the athlete's foot and the environment as shown for a variety of aspects like comfort, protection, traction, and stability (Hennig & Sterzing, 2010). However, its performance influence during kicking is unique. Soccer shoes seem to be rather generally obstructive than supportive during maximum full instep kicks. Players, who were able to neglect pain when performing maximum full instep kicks barefoot, achieved slightly higher ball velocity compared to kicking shod (Sterzing & Hennig, 2008). The mechanism to support this finding was first referred to as "forceful plantarflexion of the *foot"* (Lees, 1993) and elaborated on by Shinkai et al. (2009). The bare foot is already fully plantarflexed at initial ball contact, in contrast to the shod foot (Figure 3). At final ball contact both conditions show full plantarflexion. This indicates increased energy dissipation in the shod condition and also suggests reduced effective mass due to a less rigid ankle structure consequently leading to decreased ball velocity. A systematic analysis of the influence of isolated shoe features on ball velocity was carried out (Sterzing & Hennig, 2008).

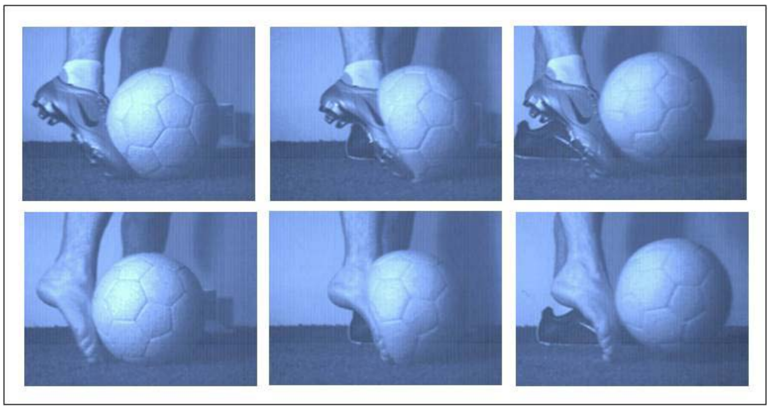

Figure 3: Ball contact plantarflexion (Sterzing & Hennig, 2008)

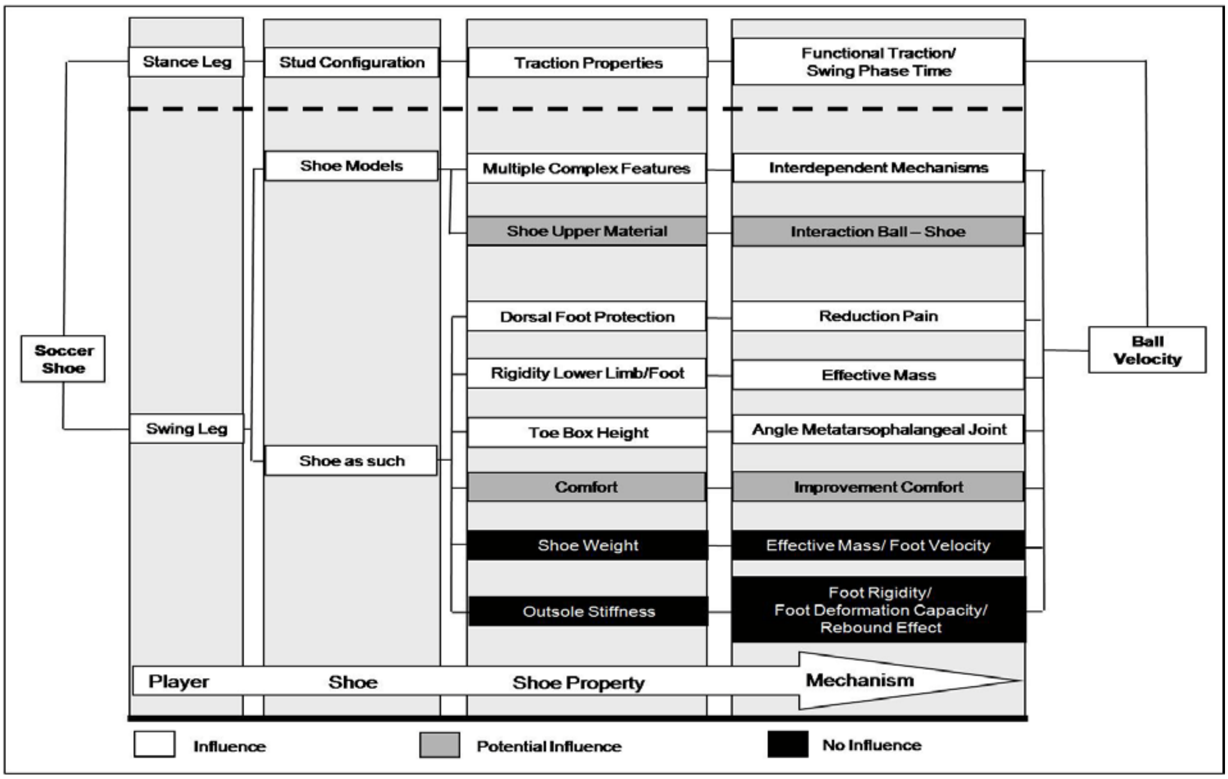

Figure 4: Shoe features and ball velocity during full instep kicks (Sterzing & Hennig, 2008)

Figure 4 provides an overview of the influencing potential of various shoe features as well as pointing out the multiple mechanisms. Finally, it needs to be acknowledged that increasing

ball velocity by soccer footwear is a complex challenge, surely influenced by stance leg traction, dorsal foot protection, the rigidity of the ankle joint and the toe box height.
The influence of the shoe on kicking accuracy has been underestimated for long. Six potential mechanisms were proposed to influence kicking accuracy: Interface pressure distribution, interface friction, spin production, foot skin sensation, functional stance leg traction, and shoe mass (Hennig & Sterzing, 2010). In contrast to kicking velocity, barefoot kicks were shown to be less accurate than shod kicks. It was speculated that the uneven, bony barefoot surface reduced accuracy of kicks compared to shod conditions, which provide a more even interface between the foot and the ball. It was concluded that avoidance of high pressure gradients across the foot is an important factor in achieving better kicking precision by footwear. However, many of the proposed mechanisms have not yet been examined.

PERSPECTIVE:

Kicking was and still is a fascinating research topic and is currently undergoing multiple new and innovative research initiatives. Interestingly, most studies have investigated kicks of stationary balls so far. Although these kicks are important and in many cases game decisive (free-kicks, penalties), the majority of passing and kicking actions take play with the ball not being stationary but rather rolling, bouncing or flying towards or away from the player in various speeds and angles, thereby increasing variability of the various kicking movements. Here, there are a lot of unsolved research issues especially with regard to timing of the kicking action that have not been examined, yet. Also, as the impact phase of the kick determines the final success of the kick, more research should focus on relating the mechanisms observed during the impact phase to the outcome with respect to speed, accuracy, spin and flight path of the ball. Thereby, whole body posture as well as specific kinematics and kinetics of the lower leg foot complex should be considered. Results of these studies should be helpful to better understand what determines the success of kicks and should be useful to give adequate advice to coaches.

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BIOMECHANICS OF BASEBALL PITCHING: IMPLICATIONS FOR INJURY AND PERFORMANCE

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All overhead throws share similar biomechanics. Baseball pitching is one of the most dynamic throws, with both high ball velocity and high rate of injury. An understanding of pitching biomechanics can help maximize performance and minimize the risk of injury (Fortenbaugh et al., 2009; Whiteley 2007). Although pitching is a continuous motion, it can be separated into a series of phases to better understand the kinetic chain. The description and the mean ± standard deviation values presented here are based upon healthy elite adult pitchers.

The objective of the windup phase is to put the pitcher in a good starting position. The windup begins when the pitcher initiates the first motion and ends with maximum knee lift of the stride leg. The pitcher typically begins with the weight evenly distributed on both feet. The stance foot then pivots to a position parallel with the rubber. The lead leg is lifted and the lead side (left side for a right handed thrower) faces the target. A pitcher will typically pitch “from the stretch” instead of from the windup when there are runners on base. When pitching from the stretch, the pitcher starts with his back foot parallel against the rubber and the front foot out a comfortable distance towards home plate. The pitcher then has an abbreviated leg lift. Forces in the upper extremity are low during this phase, thus shoulder and elbow injuries do not occur during this phase (Dun et al., 2008a).

In the stride phase, the pitcher strides his front leg (left leg for a right-handed thrower) towards the target. At the same time, the athlete separates his hands and swings them down, apart, and up. The coordination of these leg and arm motions is critical to enable optimal timing in the later throwing phases. At the time of front foot contact, the stride length should be 83±4% of body height and the lead knee should be flexed 45±9°. Also at this time, the pelvis should be 33±10° open to the target, but the shoulder line should be about 15° closed. Abduction of the throwing shoulder should be 93±11°. The elbow is flexed 90±15°, and the shoulder has 56±22° of external rotation. (External rotation is defined as 0 when the forearm is horizontal and 90° when the forearm is vertical.)

The arm cocking phase begins at the time of front foot contact. During this phase the pelvis and then upper trunk rotate to face the target while the throwing arm cocks back. The non-throwing arm is tucked in near the trunk in order to decrease inertia and increase velocity of the upper trunk rotation. The lag between pelvis rotation and upper trunk rotation is critical for generating energy from the trunk that is passed along to the throwing arm. Without proper timing of pelvis and upper trunk rotation, the athlete may have low ball speed and/or excessive loads in the shoulder and elbow (Stodden et al., 2005; Aguinaldo et al., 2007; Wight et al., 2004).

The arm cocking phase ends with the throwing shoulder in maximum external rotation (MER). MER is 181±8°; in other words, the forearm is approximately perpendicular to the trunk and the palm of the hand is facing up. Achieving such external rotation is strongly related to ball velocity. An athlete must cock his arm back far in order to accelerate his hand forward. Measured MER is not just rotation within the shoulder joint, but actually a combination of glenohumeral rotation, scapula motion, and arching of the back (Miyashita et al., 2010)

While MER is vital for ball speed, it is also a position of potential injury. In this position the rotator cuff muscles on the back of the shoulder (especially the infraspinatus muscle) may become pinched in the shoulder joint. When this muscle is impinged, it may tear during the forceful shoulder rotation. At the same time, the front of the shoulder capsule is under tension and may tear. The torques at the shoulder and elbow both peak near the time of MER, as the joints must stop the arm cocking and initiate the forward rotation of the arm. Peak elbow varus torque is 99±17 Nm. (This is about equivalent to holding a 25 kg mass in the thrower's hand at this instant.) Repetition of this varus torque can lead to tension and tearing in the elbow's ulnar collateral ligament, and as well as bone spurs in the lateral and posteromedial elbow (Dun et al., 2008b).

From this cocked position, the athlete initiates arm acceleration. Elbow extension velocity reaches 2450±250°/s and shoulder internal rotation velocity reaches an incredible 7500±900°/s. This is the fastest joint rotation documented in any sport. The biceps muscle of the upper arm contracts to decelerate the elbow extension. This contraction in the arm cocking and arm acceleration phases may lead to a tear of the shoulder labrum.

The arm acceleration phase ends with ball release. At the time of ball release, the front knee is flexed 35±12°. The front knee is extending through ball release, which allows the athlete to stop the forward motion of his pelvis and transfer energy up his body to the ball. The trunk is tilted 36±7° forward and 23±10° to the side. The throwing shoulder is abducted 94±8° (that is, the throwing elbow is approximately on the imaginary line passing through both shoulders). If the shoulder is abducted significantly more or less than 90°, there can be misalignment in the shoulder leading to damage to the shoulder capsule and surrounding tissue. Different pitchers in various throwing situations may alter the sideways tilt of their trunk; however, the shoulder abduction at ball release should always be approximately 90° (Matsuo et al., 2002; 2006).

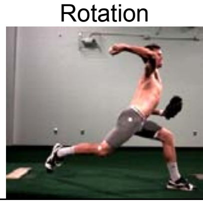
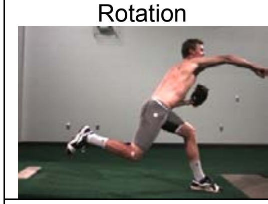
The rapid rotations of the upper trunk and throwing arm create a large force at both the shoulder and elbow. At the time of ball release, more than 1100 N are produced at both the shoulder and elbow to resist distraction. In other words, the body rotation creates forces greater than body weight that are trying to pull the arm out at the shoulder and elbow joint. Tension on the ligaments and muscles – especially the rotator cuff – may lead to tensile tears from repetitive throwing.

After ball release, the throwing arm continues to internally rotate, leaving the forearm in a pronated position. Pronation after release happens in all overhead throws – straight throws, curveballs, etc. The arm horizontally adducts in front of the chest. The trunk continues to tilt forward and the back leg steps forward. A pitcher with an abbreviated deceleration and follow-through may not be using his body to dissipate the energy produced in throwing; this may lead to excessive force in the shoulder and elbow.

A summary of proper pitching mechanics is shown in the table below. Variations from proper kinematics have been correlated with decreased ball velocity (Stodden et al., 2005) and increased shoulder and elbow kinetics (Aguinaldo et al., 2007; Matsuo et al., 2002; 2006; Wight et al., 2004) The consequences of these correlations are summarized in the table as well.

Other issues that have been shown to relate to pitching biomechanics are level (youth to professional) (Fleisig et al., 2006; 2009; Sabick et al., 2005, Ishida et al., 2006) fatigue (Escamilla et al., 2007), pitch type (fastball, curveball, etc.) (Nissen et al., 2009; Dun et al., 2008b, Fleisig et al., 2006) and technique (windup vs. stretch) (Dun et al., 2008a).

Table 1. Summary of proper and improper pitching mechanics

Phase / Event	Proper Mechanics	Pathomechanics → Consequences
Windup	Lift front leg.	
Maximum Knee Height 	Pitcher is balanced.	
Stride	Front leg goes down and forward. Arms separate, swing down, and up.	↓ push off rubber → ↓ball velocity
Foot Contact 	Front foot is planted slightly to 3B side (for a right-handed pitcher). Front foot is pointed slightly inward. Shoulder is abducted approx. 90°, with approx. 60° of external rotation.	↓stride length → ↓ball velocity Front foot open (position or angle) → ↑ shoulder and elbow force Improper shoulder external rotation and shoulder abduction → ↑ shoulder and elbow kinetics Excessive shoulder external rotation → ↓ ball velocity ↓ shoulder horizontal abduction → ↓ ball velocity
Arm Cocking	Pelvis rotation, followed by upper trunk rotation. Shoulder externally rotates, and trunk arches.	Early pelvis rotation → ↓ ball velocity Late pelvis rotation → ↑ shoulder and elbow kinetics ↓ pelvis rotation velocity → ↓ ball velocity Poor timing between pelvis rotation and upper trunk rotation → ↓ ball velocity Poor timing between pelvis rotation and upper trunk rotation → ↑ shoulder internal rotation torque
Maximum External Rotation 	Shoulder external rotation is approx. 180°. Elbow flexion is approx. 90°	↓ shoulder external rotation → ↓ ball velocity Excessive shoulder horizontal adduction & elbow flexion → ↑ shoulder kinetics
Arm Acceleration	Elbow extends, followed by shoulder internal rotation. Front knee extends.	
Ball Release 	The throwing shoulder is abducted approx. 90°	↑knee extension velocity →↑ ball velocity Improper shoulder abduction → ↓ball velocity Improper shoulder abduction → ↑elbow varus torque ↓forward trunk tilt →↓ball velocity
Arm Deceleration	Shoulder internally rotates and front knee extension continues. Trunk tilts forward.	
Maximum Internal Rotation 	Shoulder external rotation is approx 0°.	
Follow Through	Arm crosses in front of body. Trunk flexes forward.	

Definition of symbols in chart: ↓ Decreased ↑ Increased → Correlates with

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PERFORMANCE AND HEALTH CONCEPTS IN ARTISTIC GYMNASTICS

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INTRODUCTION: Artistic gymnastics attracts a large number of children and offers a range of participation levels. Gymnastics starts for many with Kindergym classes which are aimed at the development of fundamental motor skills in a game-like gymnastics environment. Most children will continue onto recreational (non competitive) gymnastics or to a nationally structured competitive program, with a select few then going onto high intensity elite training and competition from as early as 6 to 8 years of age. Considerable variability is evident in the ability of gymnasts to perform each of the apparatus during competition at all levels (inter-club, state, national, international) often due to the individual variation in physical attributes. Gymnastics requires explosive sprinting, jumping, pushing and pulling skills, together with balance and artistry on four apparatus for women (beam, uneven bars, floor, vault) and five for men (high bar, parallel bars, pommel horse, floor, vault). In competition these apparatus routines are judged subjectively by a panel of judges to identify the content and difficulty (D score) and the perfection in execution (E score).

THE CODE OF POINTS FROM A BIOMECHANICS PERSPECTIVE: The sport of gymnastics is governed by the Fédération Internationale de Gymnastique (FIG), with rules outlined for each Olympic cycle by the Code of Points (COP). The general rules and skill classifications outlined in the COP can greatly influence routine composition and therefore performance, particularly at the elite levels of gymnastics where coaches and gymnasts seek to develop a high D score. The COP can also influence the safety of the gymnasts across all levels of competition gymnastics. A recent change to the COP, for example, now requires women to stick their tumbling two foot landings on floor, without a step backwards. The COP across all apparatus for both women and men also stipulates that the feet must be held together, side-by-side when landing, with deductions for a visible gap between the feet (0.1), a step or hop (0.3), a deep squat (0.5), or a fall (1.00; FIG, 2009, p 15).

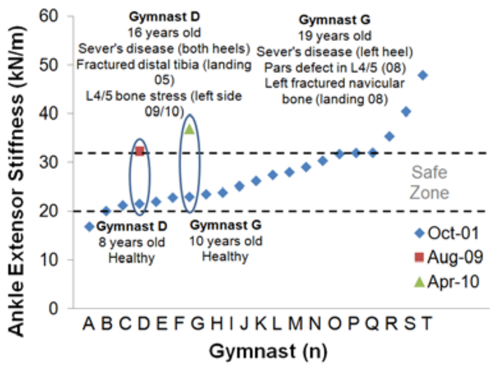
The ground reaction forces of two foot landings in gymnastics are significant during training (~5 BW) and competition (~11 BW), especially if the landing is uneven (~18 BW; Panzer *et al.*, 1988) or if there is unusual foot placement. Sticking the landing after high airborne skills requires tremendous stabilisation and eccentric strength to prevent the knee joints from collapsing (Kerin, 2006) due to the high external knee joint loads. An uneven landing or unusual foot placement can result in, for example, increased dynamic valgus knee moments (knee rotated inwards), and increases the load on the anterior cruciate ligament several-fold (Hewett *et al.*, 2005). The neuromuscular control of the centre of mass trajectory to control the body's momentum and angular rotation is also specific to the landing task, whether landing from a height without any angular rotation, or from forward or backward salto(s) (McNitt-Gray *et al.*, 2001). Wider research from other sports (e.g. netball, basketball, volleyball) has revealed that the amount of knee flexion has a greater influence on the magnitude of the impact force, than the height of the drop (Stacoff *et al.*, 1988). Ideally two foot (toe-heel) landings should be performed with even distribution of forces between the feet that are spaced roughly shoulder width apart, with actively controlled hip, knee, and ankle (plantar-) flexion with the knees over the toes (e.g. Tillman *et al.*, 2004). However the ground reaction forces in a controlled staggered landing (one foot forward) where the feet land sequentially is about half that of a traditional two foot landing. Unpublished data of 39 elite netball players landing from a high pass catch revealed landing forces of 8.2 BW (SD=$\pm$0.72) for a double foot placement, and 4.9 BW (SD=$\pm$0.13) for a staggered foot placement where one foot lands sequentially in front of the other. Whilst the recent change to the COP specifically for women's floor tumbling landings is unfortunate from a coach and gymnasts perspective (Black, 2009), greater biomechanical exploration is required to determine whether an additional step after a two foot landing does produce lower ground reaction forces. Any change to the COP must be done with caution in order to maximise safety for the competitors and to remain sensitive to the history and tradition of gymnastics.

POSITIVE AND NEGATIVE EFFECTS OF IMPACT IN GYMNASTICS: The skills necessary to compete in gymnastics require both upper and lower limb weight bearing. There are both negative and positive effects of this weight bearing impact.

The extreme forces placed on the gymnasts' body in combination with the repetitive movements and high training hours are more than likely a major factor behind the reportedly high incidence of injuries (Lilley, 2006). Although the types of injuries sustained in gymnastics are comparable with many other sports, gymnastics is unique in that the gymnasts receive the majority of their training during their childhood years (Sands, 2000). The lower extremities have been identified as the most likely anatomical location to sustain injuries (72%). The ankle has the highest reported injuries with 48% occurring during landings and 36% during take-off. The upper limbs, followed by the spine and trunk are the next most frequent sites of injury (Kirilanis *et al.*, 2003). During training, a preferred leg and hand often develops when aiming to achieve performance consistency and reliability of a skill. This can lead to a potential functional imbalance between the limbs. In non-elite competitive gymnasts (National Levels 4-6), Lilley *et al.* (2007) identified only two gymnasts out of 15 who had functionally symmetrical landings (less than 10% difference between limbs; Grace, 1985) with one gymnast having a staggering 73% of asymmetry ($\overline{X}$ = 18.14$\pm$20.46%). Recent unpublished results of 25 international development stream (elite) gymnasts performing drop landings from heights of 70 and 95 cm revealed a more favourable result, with 11 of the 25 gymnasts displaying functionally symmetrical landings ($\overline{X}$ = 6.85$\pm$14.59%, Max=32.74%) and much lower overall levels of asymmetry. Attenuating more force on one leg amplifies the risk to gymnasts of sustaining a chronic overuse injury, and is one of the most common means of injury in other sports (Kovacs *et al.*, 1999). Reduced knee flexion (Stacoff *et al.*, 1988), unusual foot placement, and increased leg stiffness (Bradshaw *et al.*, 2006; Butler *et al.*, 2003) are other potential contributing factors.

Retrospective analyses of ankle stiffness measures of self-paced double legged hopping by Bradshaw *et al.*, (2006) revealed a potential safe zone for ankle injury. As shown in Figure 1, the gymnast (A) with low stiffness had previously suffered a landing ankle injury, and four gymnasts (P, R, S, T) with high stiffness had medical histories of take-off ankle injuries such as Achilles tendonitis. A recent unpublished follow-up of two of the gymnasts (D, G) from the original study, who were still training eight years later, revealed that their ankle extensor stiffness had increased by 10.78 and 13.93 kN/m respectively. Both of the gymnasts reported Sever's disease (calcaneal apophysitis due to overuse and repetitive microtrauma of growth plates of the calcaneus in the heel) in one or both heels, and lower lumbar spine stress such as pars defect. The altered ankle extensor stiffness of these two gymnasts in combination with the reported injuries indicates that this biomechanical test has potential for monitoring injury risk, especially during peak growth years, and requires further research. Achilles strain measures using ultrasonography during a ramped protocol (three different loads) of isometric calf raises (Bryant *et al.*, 2008) may provide a further potential screening tool in gymnastics populations.

Figure 1. Ankle extensor stiffness measured from a series of six double legged hops with the knees held in an extended position from Bradshaw *et al*. (2006), with follow-up data on gymnast D and G.



Whilst there is considerable focus upon the extreme impact forces in gymnastics, the broad spectrum of loading through gymnastics participation also has many positive effects that include increased bone mass and reduced risk of osteoporosis later in life. Highly active adolescent gymnasts (n=25) have been recently shown by Greene *et al.* (2009) to display greater trabecular density, trabecular area, and bone strength at the 4% distal tibial site ($p<0.001$) and the distal radius ($p<0.001$), when compared to aged matched track and field athletes (n=34), water polo athletes (n=30), and less active controls (n=28) (Greene *et al.*, 2009). The trabecular bone tissue (the porous, spongy bone) is believed to be the most responsive to strain through loading activity (Huickes *et al.*, 2000). However despite the much lower training hours of track and field athletes ($\overline{X} = 8.4 \pm 3.9$ hrs) when compared to gymnasts ($\overline{X} = 33.2 \pm 2.3$ hrs), those athletes were superior to gymnasts for tibial cortical (the solid, outer layer of compact bone tissue) bone area and strength (measured at 14, 38, and 66% distal site), and may therefore have a correspondingly lower stress fracture risk (Wachter *et al.*, 2002). These skeletal responses may be due in part to the significantly differing overall calf size (girth), and also the differing sports surfaces for training and competition in gymnastics and athletics. Some cross-training on athletic surfaces such as running and targeting drills for vaulting, and general physical training may advantage a gymnast by increasing their cortical bone strength and reducing their overall fracture risk.

BIOMECHANICAL INFLUENCES ON TRAINING: Monitoring training objectively, with the exception of qualitative video feedback, in gymnastics is not routinely part of the sport with few known exceptions such as the Men's National Training Centre in Germany (Nissenen, 2007). This differs to other sports such as track and field and swimming where access to an instrumented track is particularly common, with swimming flumes and instrumented pools also available. Bradshaw *et al.* (2009) developed a vault timing system that measured the approach velocity through to beat board contact reliably during regular vault training of elite gymnasts. Pre-flight and table contact time were revealed to be not reliable in a training situation, but may be more reliable when used as part of monthly control (pure repeat) testing. Other researchers have attempted to instrument the apparatus such as the men's rings and high bar (e.g. Sands *et al.*, 2006) however it is unknown whether these scientific tools are used regularly in training.

Often the most important test of biomechanical research is whether it eventually improves performance and/or influences coaching (and training) practice through increased knowledge. One study that achieved success in Australia was that on target-directed running in vaulting (Bradshaw, 2004). It revealed that gymnasts' who want to perform more advanced vaults, need to be able to target and adjust their strides early in the approach in order to hit the beat board with high velocity. This is similar to the run-up skills required in long jumping (Bradshaw & Aisbett, 2006). A quote below provides a summary on how the study impacted gymnastics in Australia from the state coaches' perspective:

> "The vault targeting study was of great use to athlete and coaches. Vaulting is about hitting a target at the greatest possible controlled speed; the best vaulters in the world have the best speed. You then look at ways to develop speed, normally strength work and running and jumping drills. As a coach you want to develop this as quickly as possible so you end up with acceptable speeds and good vaults but there is always a nagging question that something is missing to get optimal performance. There are many cases where the athlete is no longer confident with their run and try to adjust where they start from. The adaptation from the training gym to podium performances where the run-up surfaces are totally different provides continual problems and need for adjustment. Enter the *Biomechanist* and a concept of targeting. We all know that the gymnast runs at a target, but have had no real idea how they do this except by the adjustment of the run up until we get something comfortable and have good speed; a long process. Liz's study of targeting in vaulting has led to a total change in the way the run and jump (take-off) is taught, and the way the significant problem of the lost run-up is viewed and dealt with. The better athletes target much earlier, the poorer and beginner ones later. This was able to be determined through non-invasive video analysis in a practical situation with a variety of level of athlete. The results from this expanded the focus of coaching vault. Drills specifically for

targeting are now included and problems with run ups are generally put into the area of an inability to target. Liz not only conducted the analysis but was also able provide drills and methodology to teach athletes to target." (Mark Calton, 11/6/2007).

Whilst there is a vast array of good quality research on gymnastics (e.g. Hiley & Yeadon, 2005; Irwin *et al.*, 2005), further fundamental research that exhibits great promise for influencing coaching (and training) practice is that of computer simulation forward modelling. This is because coaches with gymnasts seeking the epitome of competition performance are always seeking to know how to push skills further, such as the number of saltos and/or twists that can be successfully performed on floor or vault. Forward modelling such as Yeadon's (2009) simulation of the aerial skiing triple and quadruple twisting performance begins to provide information on the take-off velocity and technique required to accomplish these feats. Eventually this method of biomechanical enquiry should begin to impact upon the upper levels of elite gymnastics training.

CONCLUSION: Research on gymnastics has predominantly been focused upon the high performance participation levels, with quantitative descriptions of specific skills (e.g. Takei *et al.*, 2000), as well as epidemiological reports on injury rates, and on the impact of committed training on growth (e.g. Caine *et al.*, 2003). From a performance perspective this fails to address fundamental issues that could be broadly applied during training. Further, from an injury prevention perspective greater focus is required on the elementary years of gymnastics when the fundamental motor skills are being formed (e.g. jumping and landing), and also on the lower to middle competitive levels that involves the bulk of participants and similar rates of injury (Kolt & Kirkby, 1999). Therefore to monitor and enhance the performance of all gymnasts, key elements that maximise technique and safety need to be identified. Biomechanics can help improve gymnastics performance and reduce injury risk to provide a positive participation and/or competition experience. Research that addresses fundamental issues in gymnastics that can influence the performance of gymnasts and/or the knowledge of the coaches (e.g. Bradshaw, 2004), or can prevent injury (e.g. Bradshaw *et al.*, 2006) has provided these avenues. A wider perspective of the sport (e.g. general articles in the Code of Points), and by using techniques and knowledge developed by other scientific disciplines (e.g. motor control, engineering) can further enhance the influence of biomechanics on this sport in the future.

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MOTOR VARIABILITY AND SKILLS MONITORING IN SPORTS

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The aim of this paper is to present a brief review of the role that movement variability plays in the analysis of sports movement and in the monitoring of the athlete's skills. Motor variability has been traditionally considered an unwanted noise to be reduced, but recent studies have revalued its role and have tried to understand whether it may contain information about the neuro-musculo-skeletal organisation. Issues concerning both views, different approaches to variability, open questions and future perspectives will be discussed.

KEY WORDS: movement variability, motor skills, performance, entropy, coordination.

INTRODUCTION:

The study of movement variability has been gaining increasing interest in the sports biomechanics community. In the last five years at the International Society of Biomechanics in Sport meetings there have been two "Dyson" lectures (Bartlett, 2005; Hamill et al., 2006), several keynote talks (e.g. Bartlett, 2004; Hamill et al., 2005; Wilson, 2009), and an applied session (at ISBS 2009), that have evidenced the importance of motor variability (MV) and coordination variability (CV) in the analysis of sports movements.
Despite the efforts spent by researchers, many issues concerning the variability of human motion are still to be thoroughly addressed and/or are waiting for comprehensive explanations. Among them, for example: the meaning of MV; the information MV may provide; the possible relations between MV and performance or between MV and the acquisition/development of motor skills. Furthermore, variability is fundamental in the definition of the experimental design and may influence the validity of the obtained results.
This paper presents a report about the role that MV and/or CV may have in the process of monitoring the athlete's biomechanical qualities. In particular, attention will be focused on the importance that movement variability has in the attempt of describing motor skills.

MONITORING ATHLETIC SKILLS: THE DUAL NATURE OF MOVEMENT VARIABILITY

Motor skills represent the ability to obtain a predetermined outcome with a high degree of certainty and maximum proficiency (Schmidt & Lee, 2005; Newell & Ranganathan, 2009). Hence, the process of learning or improving sports skills involves the capability of producing a stable performance under different conditions: only repeated motor performance reflects mastery in carrying out a desired task.
According to this definition and thinking of some of the biomechanists' principal aims – e.g. describing the athlete's kinetic and kinematic peculiarities, evaluating the correctness and proficiency of their movement, preventing possible injuries – MV may emerge as an unwanted drawback that should be eliminated or at least reduced. In fact, when trying to capture the biomechanics of individual technique, research should depict the core strategy that governs the movement, regardless of the variations that emerge across repetitions.
Nevertheless, MV always occurs when the same action is repeated and even the elite athlete cannot reproduce identical motor patterns. MV is inherently present throughout the multiple levels of movement organisation. It remarkably manifests not only between but even within individuals and may be associated to the extreme complexity of the neuro-musculo-skeletal system and to the redundancy of its degrees of freedom (e.g. Newell et al., 2006; Bartlett et al., 2007). MV may not correspond only to randomness but also to functional changes whose investigation might unveil information about the system health, about its evolutions, and about its flexibility and adaptability to variable external conditions (Hamill et al., 1999; Bartlett et al., 2007).

Therefore MV possesses a dual connotation. (1) It is an impediment that obstacles a straight description of the actual individual status through standard approaches. Moreover, it hinders the detection of the small inter-individual differences or intra-individual changes that often characterise the sports field. At the same time, (2) it is the reflection of inherent proprieties of the neuromuscular system and may contain important information that should not be neglected.

MOVEMENT VARIABILITY AS NOISE

According to the conventional control theory approach, movement variability is made equal to noise (Equation [1]) that prevents the final output from matching the planned program (Bartlett et al., 2007; Bays et al., 2007). The noise may corrupt the different levels of motor organisation (V_{eb}, i.e. errors in the sensory information and in the motor output commands) and may be caused by the changeable environmental conditions (V_{ee}) or by measuring and data processing procedures (V_{em}).

[1] $V_e = V_{eb} + V_{ee} + V_{em}$

This view of MV has important implications for the investigation of sports skills and leads to the need for proper experimental designs and data reduction (Bartlett et al., 2007). MV should be assessed before proceeding with any kind of biomechanical assessment and a full analysis of an individual's motor behaviour should involve the evaluation of an appropriate number of repetitions (e.g. Bates et al., 1992; Rodano & Squadrone, 2002; James et al., 2007; Preatoni, 2007). The selection of a single representative trial may be arbitrary and results derived from analyses of such performances may be misleading (Bates et al., 1992). While several researchers have thoroughly investigated the reliability of normal walking variables, relatively fewer studies have been conducted to assess the variability of kinematics and kinetics in sports movements (Preatoni, 2007). This lack is amplified by the huge variety of motor tasks that are performed by athletes in many different sports disciplines. The effective trial size needed to depict a representative biomechanical behaviour likely depends on the activity, on the subject and on the variable under investigation (e.g. Bates et al., 1992; Rodano & Squadrone, 2002; James et al., 2007; Preatoni, 2007). For example, by studying race walking and vertical jump exercises in a population of high-level athletes, Preatoni (2007) and Rodano & Squadrone (2002) found that 11–16 repetitions were necessary to obtain stable estimates for kinematic and kinetic parameter.
The use of suitable statistics is also necessary to obtain a meaningful summary of the collected parameters or curves, in order to discover the most typical features and to predict whether a pattern is representative for the athlete's skill description or not. Non-parametric estimates of central tendency and spread appear to be more robust to the presence of outliers (Chau et al., 2005; Preatoni, 2007).

MOVEMENT VARIABILITY AS INFORMATION

Recent investigations have supported the idea that inter-trial variability (V_{tot}) does not correspond to noise only, but is a combination (Equation [2]) of random fluctuations (i.e. error, V_e – Equation [1]) and functional changes that may be associated with properties of the neuromotor system (V_{nl}) (e.g. Hamill et al., 1999; Bartlett et al., 2007):

[2] $V_{tot} = V_e + V_{nl}$

V_{nl} may be interpreted as the flexibility of the system to explore different strategies to find the most proficient one among many available. This flexibility allows for learning a new movement or adjusting the already known one by gradually selecting the most appropriate pattern for the actual task (e.g. Dingwell et al., 2001; Riley & Turvey, 2002). The subject is thus able to gradually release the degrees of freedom that have been initially frozen to gain a

greater control over an unfamiliar situation (e.g. Hamill et al., 2005; Newell et al., 2006). Changes in the contributions of V_e and V_{nl} to the total variability may be related to changes in motor strategies and may thus reveal the effects of adaptations, pathologies and skills learning (e.g. Dingwell et al., 2001; Bartlett et al., 2007).
The conventional approaches to the issue of MV only quantify the overall variability, but are not effective in evaluating the information MV conveys. The use of nonlinear dynamics tools (e.g. entropy measures) or the analysis of coordinative features (e.g. continuous relative phase) represent alternative instruments to explore the nature of motion variability or its relation with performances, skills or injury factors. Only recently and only a few authors have used these methods to investigate MV in sports and in elite athletes in particular.
Preatoni et al. (2007; 2010-under review) studied the nature of MV by measuring sample entropy in kinematic and kinetic variables during race walking. Their results confirmed that MV is not only random noise but also contains information about the neuromuscular organisation. Possible changes in variability across the individual's testing session were also analysed. Results suggested that the structure of variability appears invariant and that no adaptation effects emerge when a proper experimental protocol is followed. Other authors have focused their attention on injury factors (e.g. Hamill et al., 1999; Hamill et al., 2005) or on coordinative patterns (e.g. Seay et al., 2006), by studying the variability in phasing relationships between different elements of the locomotor system (body segments or joints). Fewer works have concentrated their attention on the relation between sports skills and MV/CV, with practical implications for performance monitoring and training purposes. Wilson et al. (2008) studied how coordination variability changes in relation with skills development in the triple jump. Preatoni et al. (2007; 2008; 2010-under review) reported different levels of entropy, in selected variables, between elite and high-level race walkers. Furthermore, Preatoni et al. (2007; 2008; 2010-under-review) and Donà et al. (2009) evidenced how advanced methodologies may be an important means for finely investigating individual peculiarities – e.g. subtle changes over time that may be due to underlying pathologies – when no apparent changes occur at a macroscopic level.

CONCLUSION

Similar performances in sporting events are often the result of different motor strategies, both within and between individuals. These subtle discrepancies are typically less detectable than the ones that emerge in clinical studies, and are often concealed by the presence of variability. Hence, the conventional use of discrete variables or continuous curves may be ineffective, while the study of movement and/or coordination variability may make important neuro-musculo-skeletal features emerge.
This paper has briefly presented the "double" role that motion variability plays in the analysis of sports movement, being concurrently a limitation and a potentiality. Regardless of the point of view from which we look at MV, many efforts are still needed to gain a more thorough insight into this issue. In fact, for example, there is still lack of: (1) reference values and database, that could help in the interpretation of MV/CV in sports; (2) knowledge of the relations between causes (e.g. detrimental behaviours, motor learning) and effects (e.g. changes in the analysed variables or indexes) (Hamill et al., 2005; Bartlett et al., 2007; Preatoni, 2007); (3) integration between the outcomes of the different methods of investigation; (4) ability in translating complex approaches and results into suitable information that may be easily read as feedback and thus applied on the field.

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ISBS 2010

Applied Sessions

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Nordic Skiing

NORDIC SKIING BIOMECHANICS AND PHYSIOLOGY

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Cross-country skiing provides unique movement patterns and unique athletes for sport science research. The integration of physiology and biomechanics into research projects has provided insights into systemic solutions regarding high intensity, whole body exercise and how workload is distributed. Newly introduced ski sprint events have stimulated continued research to better understand performance and human capacity limitations.

KEY WORDS: aerobic capacity, cross-country skiing, energy cost, workload distribution

INTRODUCTION: In the interdisciplinary field that we today refer to as kinesiology or exercise science, there is a long historical connection to researchers in Scandinavia who used familiar modes of activity such as running, cycling and skiing in their physiological and biomechanical experiments. Early in the 20th century, field measurements of metabolic costs were documented for a variety of forms of locomotion including cross-country skiing (e.g. Liljestrand & Stenström, 1920). Later, Christensen and Högberg (1950) documented an oxygen uptake of 5.2 $L{\cdot}min^{-1}$ in a skier. More recently, the widely read *Textbook of Work Physiology* by Åstrand and colleagues (original edition in 1970) included an illustration of a skier with a Douglas bag on his back during oxygen uptake measurements while cross-country skiing. These and many other examples from Scandinavian researchers form an important core in the entire exercise science literature and included measurements from skiers not for convenience but because of the unusual physical capacities of the athletes involved in ski racing, the unique quadripedal characteristics of the movement patterns of skiing, and because understanding physiological and biomechanical responses in skiing can teach us much about human physiology, mechanics, and motor control more generally. It is in that spirit that this symposium presents an interdisciplinary perspective on "Nordic" skiing—an activity that has evolved considerably over the decades but which continues to provide insights into human systems.

While ski artifacts date back several millennia in northern regions with long winters, it is over the last few centuries that substantial changes have been made to ski design (Figure 1). Formenti et al. (2005) have documented the evolution of ski construction and re-created representative models of skis for usage testing under controlled conditions. Their testing showed the substantial decrease in metabolic cost per meter of skiing that has been achieved through equipment adaptations (Figure 2). This is partly due to reductions in the mass of the equipment (about 6 kg vs. 2 kg for modern skis, bindings, boots and poles) but perhaps primarily due to reductions of the coefficient of friction of skis on snow (μ of about 0.05 vs. 0.01 for modern skis on a packed snow surface, Formenti et al., 2005). These measurements were made under controlled snow conditions. In typical loose snow field conditions, it is likely that both frictional forces and metabolic costs would be substantially greater. This can be observed in the work of Christensen and Högberg (1950) where the results can be manipulated to determine metabolic cost per meter of travel (about 5 J/kg/m compared with about 3 J/kg/m measured by Formenti et al. for a packed and mechanically prepared snow surface).

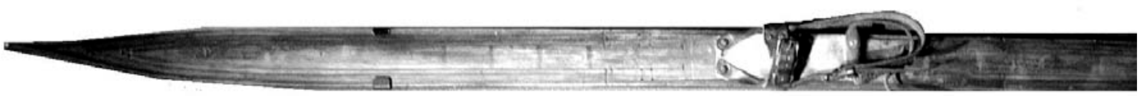

Figure 1. By the late 1800s, ski equipment had evolved from the rudimentary wooden planks of previous millennia to more refined and lighter structures. This example from the Oslo Ski Museum was thinly carved and used well developed leather bindings, but was several times the weight of modern skis.

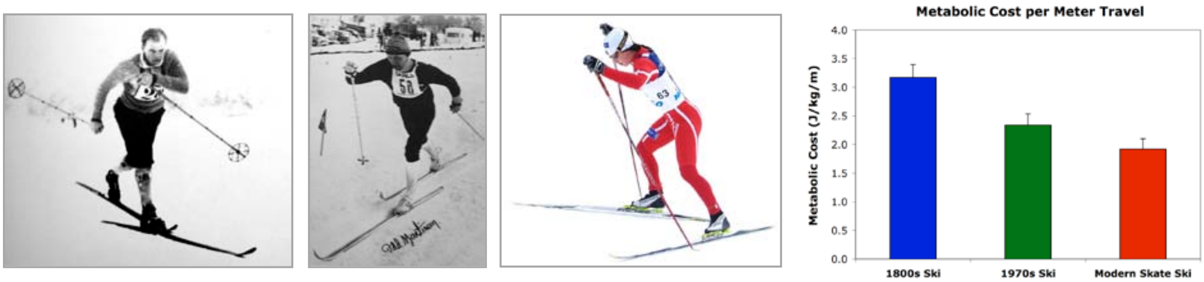

Figure 2. Equipment, ski techniques, and metabolic costs have evolved over the past century. The photos show Norwegian national and Olympic champions from the 1920s (Torleif Haug), from the 1970s (Odd Martinsen) and the 2000s (Marit Bjørgen). The relatively heavy wooden skis that Haug used were supplanted by lighter wood laminates that Martinsen raced on and later by fiberglass skis which could be optimized for skating (shown in the photo) or for classical techniques similar to Haug or Martinsen's technique of earlier years. Metabolic cost per meter of skiing on such equipment has systematically decreased over time. Photos of Haug and Martinsen were adapted from displays at the Oslo Ski Museum. Metabolic cost data are from Formenti et al. (2005).

Much of the early research using cross-country skiers was aimed at achieving better understanding of the physiological characteristics of athletes and the demands of various forms of exercise. As the sport itself has evolved, considerably more attention has been focused on technical aspects of ski performance. The new skating techniques which were introduced about 1985 have continued to evolve and provide a challenge to optimization of race performance. One emphasis of recent biomechanical studies has been to better understand the differing whole body workload distributions of various ski techniques. Studies of the interaction of human physiology with the mechanical demands of varying techniques has enhanced our understanding of how human systems respond to competing requirements from upper and lower extremities during high intensity ski racing. Recent biomechanical and physiological research lines are described below.

BIOMECHANICS OF NORDIC SKIING: Biomechanics research in sport often follows a trajectory from simple kinematic studies during early years to more complex three-dimensional kinematic analysis and finally to kinetic analyses involving measurement of force or determination of mechanical energy characteristics. Ski biomechanics research has paralleled this path in some respects but has also diversified in unexpected ways as the sport has evolved new techniques and new sprint events. In recent papers one can find examples of rather simple two-dimensional kinematic analysis combined with complex physiological and kinetic measurements integrated together (e.g. Zory et al. 2010). The complexity of field measurements has often demanded some simplification to the instrumentation. Few studies (Smith, 2000) have managed three-dimensional motion analysis combined with force measurement on snow. The result of this technical challenge is that we have very poor estimations for the various ski skating techniques regarding the components of force generated by skis and poles on snow.

Laboratory studies using rollerskis on large treadmills and full motion analysis systems have in part reduced this shortcoming. Under such controlled conditions, comparisons of skating techniques can be made which help explain technique choices for submaximal speeds (Figure 3). Unfortunately, sprint ski skating (Figure 4) has been too challenging to attempt on a treadmill, so we cannot explain in force component terms the observed performance characteristics.

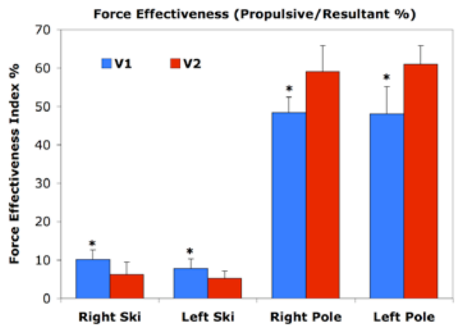

Figure 3. Force effectiveness index for skis and poles on a 5° slope. V1 ski forces were more effective than V2 while V1 pole forces were less effective ($p<0.05$). Data from Smith et al. (2009).

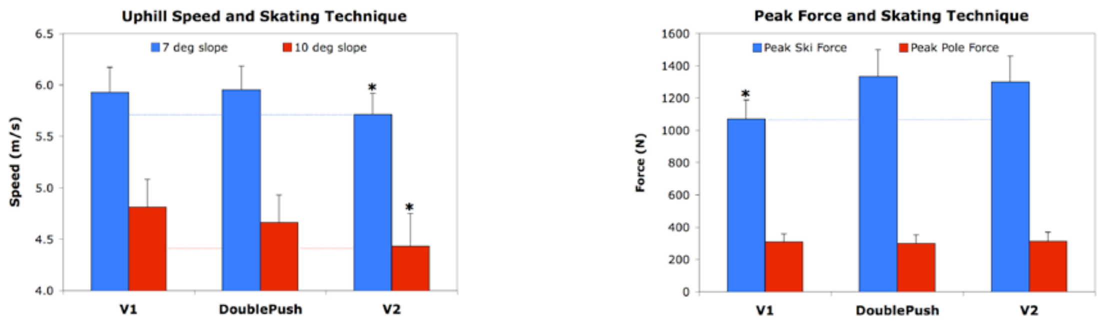

Figure 4. Uphill sprint speed and peak forces for commonly used V1 and V2 skating techniques and a newly developed doublepush technique. On uphill terrain, the V2 technique was slower and required generation of greater skating forces than V1. The new double-push technique was equally fast as V1 but required a complex ski hop maneuver and higher skating forces. Data are from Stöggl et al. (2010). * $p < .05$

PHYSIOLOGY OF NORDIC SKIING. Cross-country skiing is one of the most demanding endurance sports. It imposes extensive physiological challenges due to the multiple changes between, and utilization of, different skiing techniques, each involving the upper and lower body to various extents. The uniqueness of the sport has contributed to significant interest from researchers in sport science due to their ambition to understand more about the limiting factors of performance. Compared to other endurance sports, cross-country skiing is a complex racing form with a comprehensive diversity of locomotion types on various types of terrains and different inclinations. This indicates that, in comparison to other endurance sports, the skier's aerobic capacity in several techniques is critical for performance.

As early as in the beginning of the 20th century, Liljestrand & Stenström (1920) carried out intensive research which included not only pulmonary oxygen uptake but also determinations of cardiac output. One of the conclusions Liljestrand and associates reached was that in subjects who were studied in several different forms of locomotion the highest oxygen uptake was always found when skiing. Maximal cardiac outputs in excess of 40 $L \cdot min^{-1}$ and stroke volumes over 200 ml have been measured in elite cross-country skiers, with maximal oxygen uptake values above 6 $L \cdot min^{-1}$ (Ekblom & Hermansen, 1968) with relative values in the range of ~80-90 $ml \cdot kg^{-1} \cdot min^{-1}$ (Bergh & Forsberg, 1992). Partitioning aerobic energy costs for various regions in the body is more difficult. Modern techniques such as PET (Positron Emission Tomography) or MRI (Magnetic Resonance Imaging) to study the metabolic response or blood flow of specific muscles can be performed at least for some muscles but only for simple and well standardized muscle contractions rather than in composite movements as in cross-country skiing.

To get a more complete picture of the energy costs, traditional techniques using measurements of regional blood flow have to be performed and combined with arterio-venous (a-v) differences over the region under study. Only in cross-country skiing exercise have investigations been completed by our research group in order to analyze metabolism and aerobic energy turnover for arms and legs separately (Calbet et al., 2004; Calbet et al., 2005). During submaximal diagonal skiing, double poling, and leg skiing, cardiac output (26-27 $L \cdot min^{-1}$), mean blood pressure (~87 mm Hg) and systemic vascular conductance, systemic oxygen delivery and pulmonary VO_2 (~4 $L \cdot min^{-1}$) attained similar values regardless of exercise mode. However, distribution of cardiac output (Figure 5) was modified depending on the musculature engaged in the specific

Figure 5. For skiers at the highest intensities, the cardiovascular system must distribute blood flow and hence oxygen delivery in a manner which avoids systemic hypotension.

skiing technique. Skeletal muscle vascular conductance was restrained during diagonal skiing to maintain the arterial pressure during exercise and avoid hypotension.

A new racing form in competitive cross-country skiing is the sprint event in which skiers perform four separate races of 1200–1800 m (racing times from 2 to 4 min), starting with an individual time-trial qualification round, and thereafter heats based on a knock-out system. A relevant question is whether aerobic capacity is also of importance in such sprint races?

In a recent study that investigated the physiological characteristics of eight world-class (WC) and eight national-class (NC) Norwegian sprint cross country skiers (Sandbakk et al., 2010), the WC skiers showed an 8% higher VO_2peak than the NC skiers. In addition they reached a VO_2 plateau time nearly twice that of NC skiers, had a higher gross efficiency and 8% higher peak speed. These results indicate that aerobic capacity can differentiate between sprint skiers of different performance levels. Moreover, efficiency and the ability to reach high skiing speed is crucial for performance supporting the importance of ski technique.

SUMMARY: Cross-country skiing is both mechanically and physiologically complex. The main freestyle and classical styles, are subdivided into nine different sub-techniques. Skiers adapt to changes of inclination and speed through changes in pole and leg kinematics, kinetics, and workload distribution which affect physiological characteristics throughout the body. To achieve a better understanding, the use of an integrative biomechanical and physiological approach is an important tool for increasing knowledge and enhancing performance.

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ISBS 2010

Applied Session

Weightlifting

BASIC PERFORMANCE CUES FOR TEACHING THE SNATCH AND CLEAN TO NON-OLYMPIC WEIGHTLIFTING ATHLETES

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OVERVIEW: The intent of this article is to introduce the basic tenets of performing the movements of Olympic Weightlifting. Precise performance of these exercises requires a multitude of physical qualities coordinated at very high rates of speed. Through my experience, many strength and conditioning coaches throughout the world utilize these lifts or their derivatives for enhancing the overall productivity of athletes under their charge. These exercises are traditionally based in the strength and power sports, but these exercises have been used with athletes who participate in endurance activities as well with very good results. Most competitive activities will have some aspect of power which can be enhanced by applying the Snatch and/or Clean in the training regimes.

The movements are very similar in their performance; the main differences being final resultant bar accelerations and velocities, hand spacing and where the bar is fixed in relation to the body at the completion of the movement. The following is a brief synopsis of the main coaching points to teach athletes at the onset of applying these movements into the training sessions.

Coaching Points: The Starting Position and pull are the same for each exercise with the exception of hand spacing on the bar and the height at which the bar will "brush" against the upper thigh or lower hip region as it accelerates upward to the receiving position. In the Snatch, the hands will be spaced approximately the width between elbows when extending the humerus laterally to a 90 degree angle in relation to the torso. The Clean will have hand spacing equal to the width of holding the hands just lateral of the hips, similar to the traditional anatomical position.

- All Body Levers Are "Tight"
- Feet Slightly Turned Out and in the "Vertical Jump" Position
- The Back Is "Flat" and Even Concave
- Arms Are Straight and the Elbows Are Out
- The Head Is Up and the Eyes Are Focused Straight Ahead
- The Hips Are Higher Than the Knees
- The Shoulders Are In Advance of the Barbell

The Pull
- The Barbell Moves Back Toward the Athlete
- The Hips and Shoulders Rise at the Same Time
- The Head Stays in a Level Position
- The 2nd Pull Must Be Faster Than the 1st Pull
- The Athlete Should Try To Stay "Flat-footed" as Long as Possible
- The Arms Bend Only To Pull the Athlete Under the Bar
- The Feet Move From a Pulling Position To a Receiving Position

Snatch

- The lifter approaches the barbell and sets the feet
- Then adopts the starting position
- Inflate chest; set back
- Shoulders are in advance of the bar
- Arms are straight
- Eyes are focused straight ahead
- Weight is distributed evenly
- It is imperative to push with the feet initially and as the barbell passes the knees acceleration should constantly increase
- The lifter then extends the body upward in a violent motion
- The shoulders shrug, the arms are straight and the weight shifts from the heels to the ball of the feet.
- The lifter will exert so much force that it will continue to rise while jumping underneath the barbell.
- After jumping under the bar, the lifter will receive the barbell at arms' length and proceed into a full squatting position and stabilize the system (bar and body) in the bottom position.
- Upon acquisition of a stable posture, the athlete will rise out of the squat into a standing position.
- In a controlled manner, the bar is then allowed to return to the floor by stepping out from beneath while maintaining hand contact with the bar to ensure its proper and safe return.

Clean

- The lifter approaches the barbell and sets the feet
- Then adopts the starting position
- Inflate chest; set back
- Shoulders are in advance of the bar
- Arms are straight
- Eyes are focused straight ahead
- Weight is distributed evenly
- It is imperative to push with the feet initially and as the barbell passes the knees acceleration should constantly increase
- The lifter then extends the body upward in a violent motion
- The shoulders shrug, the arms are straight and the weight shifts from the heels to the ball of the feet
- After the lifter finishes the pull, the athlete pulls under the bar and catches it in the receiving position
- The bar should rest across the shoulders and clavicles while keeping the chest and elbows elevated, while moving into a full squatting position.
- When a stable posture is established, the athlete will rise to the standing position and then return the bar to the floor safely.

PRACTICAL APPLICATION: These are the basic principles involving performance of the two movements. If applied correctly, the exercises can be included in training regiments of all sports to enhance the various physical qualities needed for a higher level of performance production in the sport. The practical portion of this presentation will demonstrate the principles, techniques and performance of the Olympic lifts.

ISBS 2010

Applied Session

Motion Analysis

ADVANCED APPLICATIONS OF MOTION ANALYSIS IN SPORTS BIOMECHANICS

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Motion analysis is utilized to capture various raw linear positional data of markers placed on a body performing a movement sequence in sport. The application of motion analysis involves biomechanical modeling to calculate various kinematic and kinetic derived variables to understand the principles of motion. Since biomechanists are recruited as consultants to teams and coaches in sport teams, it is essential that they apply motion analysis to not only evaluate individual performance, but suggest methods of optimizing technique for enhanced performance and injury risk reduction. This presentation will show how advanced applications of motion analysis can lead to the biomechanical enhancement of sports performance.

KEYWORDS: motion analysis, kinematics, kinetics, performance analysis, talent identification.

INTRODUCTION & OVERVIEW:

A detailed understanding of the biomechanics of human motion in sports generally requires the service of a multiple camera three-dimensional motion analysis system to film, capture, track, digitize and analyze motion over time. A variety of motion analysis capture methods such as optical, electromagnetic and image-based techniques can be used. However, they all serve a common objective to obtain raw positional data of segment points that can be filtered and used to calculate various kinematic and kinetic derived variables. These variables are applied to quantify and experimentally validate descriptions of sports technique, and also provide biomechanical explanations of the motion patterns observed in sports.

If aided by a qualitative analysis, then coaches can use biomechanical descriptors of technique to improve the quality and clarity of teaching instruction (Knudson, 2007). In addition, a biomechanical understanding of sports technique potentially leads to the optimization of technique with respect to various performance outcomes for different sets of constraint functions (Hatze, 1983; Gonzalez, 1989). Hence, the aim of this presentation is to show how the advanced application of motion analysis in sports has the potential to significantly improve the performance of athletes.

EXTENDING MOTION ANALYSIS METHODOLGY:

Apart from performing the standard preliminary methodology for using a motion analysis system, such as defining the capture volume, synchronizing force platform data with the cameras, completing calibrations and developing appropriate marker systems, there are a number of other external procedures that need to be completed for sports biomechanical analysis:

1. Export motion analysis positional data into a numeric computational software such as Matlab (The Mathworks, Inc.) or symbolic manipulation software such as Maple (MapleSoft, Inc.) or Mathematica (WolframResearch, Inc.).
2. Use an algorithm to automatically calculate the cut-off frequency and smooth data with appropriate filter, such as fourth-order low pass Butterworth filter.
3. Write algorithms that will automate the full range of kinematic calculations such as:
    a. Relative temporal differences of linear and angular velocity maxima to quantify segmental sequencing (Ferdinands et al., in review)
    b. Three-dimension rotation or angular velocity vectors to calculate the segmental planes of motion

   c. Angular velocity vectors and moment arms from axes of rotation to end-effector to calculate segment velocity contributions (Sprigings et al., 1994).
   d. Derivation of segment interaction equations and calculation of interactive moments (Putnam, 1993).
4. Use software to calculate inverse dynamics such as Kintrak (University of Calgary), KinTools RT (MotionAnalysis), Vicon BodyBuilder (Vicon), Mathematical Mechanical Systems Pack (Wolfram Research, Inc.), etc
5. Export inverse dynamics into numeric computational software or symbolic manipulation software to perform further kinetic analysis such as:
   a. Joint and muscle powers
   b. Segment interaction analysis
   c. Power flows
6. Create multi-segment forward solution model in human simulation software such as MADYMO (TASS, Inc.), LifeMOD (LifeModeler, Inc.), OpenSIM, Mechanical Systems Pack (Wolfram Research, Inc.). Export inverse dynamic data into forward solution model.

SPORTS BIOMECHANICS ANALYSIS

Motion phase characteristics: Traditionally, in sports biomechanics a motion is divided into phases. There are four generic phases of motion: backswing, transition, downswing and follow-through. Each phase has its own defining mechanical characteristics: the backswing phase creates the end-effector arc length, the transition phase activates the stretch shortening cycle, the downswing phase accelerates the end-effector, and the follow-through phase slows down the end-effector. Motion analysis procedures need to be applied to precisely define and evaluate the biomechanical characteristics of each phase of motion that determine the overall performance outcomes of the motion.

Planes of motion: In sports, body segments are invariably forced to move in different planes of motion. In some field studies, these planes of motion are determined by projecting vectors on two-dimensional planes. However, such analyses are subject to projection errors. To gain an accurate assessment of global and relative planes of motion, three-dimensional angular rotation and velocity vectors have to be calculated. This is particularly important for sports such as golf, which place a coaching emphasis on shoulder, hip, arm and shaft planes.

Segmental sequencing: Angular velocity vectors are calculated to determine the segmental sequencing patterns in athletic motion. A general adherence to the proximal to distal sequencing scheme promotes effective performance in most sports that produce high-end effector velocities, such as in the golf swing, cricket bowling and baseball pitching. However, there are a number of violations in this classical sequencing scheme, such as the timing of long axis rotation in the tennis serve (Marshall and Elliott, 2000). Sequencing patterns are particular to each sport, and may also differentiate between elite and amateur athletes.

Kinetic analysis: Although sequencing patterns can be observed and quantified, there is no account of the causal mechanisms underlying these patterns without a kinetic analysis. For instance, it has been shown that the shank lags behind the thigh during the early swing phase in kicking due largely to the interactive moment resulting from the forward acceleration of the thigh (Putnam, 1993). The identification of the causal mechanisms of movements requires a combination of joint torque, power flow and segment interaction analysis.

Summation of segmental velocities: In the classic kinetic link principle, each succeeding distal segment is activated after the corresponding proximal segment has reached its maximum linear or angular velocity (Marshall and Elliott, 2000). Such a scheme describes the sequential summation of segmental velocities: the maximum velocity of the proximal segment or joint is added to its corresponding distal segment throughout the kinematic chain. This scheme does not occur ideally in actual motion sequences. However, mathematically, certain percentages of segmental velocities, both linear and angular, are added throughout the kinetic link chain.

Stretch-shortening cycle activation: Physiologically, it is well-established that the pre-stretching of muscles increases the strength of the subsequent concentric contraction. Stretch shortening cycles are activated at various times in sports. However, there is a distinct phase known as the transition phase in which the stretch shortening cycles of the major power actuating muscles are most strongly activated during eccentric contractions. Motion analysis techniques need to identify the various stretch shortening cycles that occur in movement patterns. Traditional approaches have used static separation angles, such as X-factor and X-factor stretch in the golf swing (Cheethan et al., 2001). Since the stretch shortening cycle is activated dynamically, it may be more effective to study accelerations and kinetics together.

PRACTICAL APPLICATIONS:

As the mechanisms of movement sequences in sport are multi-layered and interdependent, biomechanists often have to apply several motion analysis technqiues to meaningfully assess sports performance. Examples that will be presented are listed below:

Baseball pitching:
- Relative temporal phase of motion differences and throwing accuracy
- Identification of sub-transition phases in pitching
- Differences in segmental sequencing in elite and amateur pitchers (Matsuo et al., 2001)
- Activation of stretch shortening cycle in terms of pelvic-shoulder separation angle
- Causal mechanisms of arm segment angular velocities through segment-interaction analysis (Hirashima et al., 2008)

Kicking:
- Activation of stretch shortening cycle in terms of thigh flexion, knee flexion, time occurrence of maximum knee flexion and knee flexion angular acceleration
- Segmental sequencing in maximal velocity instep kicking (Nunome et al., 2002)
- Causal mechanisms of proximal to distal sequencing in kicking (Putnam, 1993)

Golf swing:
- Identification of swing plane types (Coleman and Rankin, 2005)
- Activation of static stretch shortening cycle in terms of X-factor and dynamic stretch shortening cycle in terms of X-factor stretch (Cheethan et al., 2001)
- Major segment velocity contributions in the golf swing (Ferdinands et al., 2004)

Cricket bowling:
- Segmental sequencing of elite fast bowlers (Ferdinands et al., in review)
- Forward solution model to reduce shoulder counter-rotation in bowlers (Ferdinands et al., 2008)
- Spinal kinetics and lumbar injury in fast bowlers (Ferdinands et al., 2009)
- Bowling legality analysis (Ferdinands et al., 2007)

Tennis serving:
- Identification of violations in segmental sequencing of the tennis serve (Marshall et al., 2000)
- Major segment velocity contributions in serving

CONCLUSIONS:

Biomechanists are becoming increasingly recruited as consultants to teams and coaches in sport teams. If they are to be successful, then it is essential that they apply motion analysis to not only evaluate individual performance, but suggest methods of optimizing technique for enhanced performance and injury risk reduction. As in all scientific fields, the most successful biomechanists will be innovative and pioneers in the technical development of their specialised sport. In this presentation, it will be shown that biomechanists can apply motion analysis to improve sports performance in the following ways:

1. To develop a detailed descriptive biomechanical analysis of sports technique
2. To establish the biomechanical criteria that are characteristic of optimal technique
3. To establish the validity of coaching intervention measures on selected performance outcomes
4. To perform a biomechanical performance blueprint or profile
5. To perform a quantitative talent identification survey

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GOLF SWING MOTION ANALYSIS: CHALLENGES AND SOLUTIONS

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With advanced motion capture technologies available, the scope and depth of motion analysis now is mainly limited by the capability of the analysis software. Golf swing analysis requires advanced motion analysis methods such as detection of the true impact instant, computation of the impact conditions, definition and use of local reference frames (body and club) in various ways, and determination and use of the functional swing plane. The purpose of this paper was to identify and present solutions for the unique challenges encountered in golf swing motion analysis and to demonstrate the application of such solutions by using Kwon3D, a comprehensive motion analysis program.

KEY WORDS: Motion analysis software, motion capture, golf swing analysis.

INTRODUCTION:

The ultimate capability of a motion analysis system is determined by two major factors: motion capture capability and scope and depth of analysis allowed by the analysis software. The motion capture capability of a system depends largely on the hardware used (cameras, image processor, and controller) and the control software. With recent advancements in the real-time motion capture technologies and availability of the real-time systems, the capability of the data analysis software has become the main limiting factor.

Computation of select meaningful kinematic and kinetic parameters based on the raw position data, the outcome of motion capture, is the primary role of the analysis program. The kinds of biomechanical analyses allowed and the types of kinematic and kinetic variables provided are limited by the capability of the analysis software. For this reason, it is not rare for investigators to develop additional study-specific software programs to overcome the limitations of the general-purpose motion analysis program attached to the motion capture system. The downside of this route, however, is that each study will require a study-specific program which may not be applicable to other types of studies. When a change is made in the analysis, the program codes must be changed accordingly.

The ultimate solution to this problem is to develop a flexible, adaptive and comprehensive motion analysis program in which the scope and depth of analysis can be expanded and the need for additional study-specific programming is practically eliminated. Kwon3D (Visol, Inc., Seoul, Korea) is an example of such a program, characterized by high flexibility, adaptability, and expandability. The notion of a comprehensive analysis program is particularly attractive as the raw data now can be transferred easily from one platform to another via the file formats commonly supported by different motion captures systems, e.g. the C3D format.

A golf swing is a complex movement that presents unique challenges to the investigators, requiring advanced motion analysis methods. For instance, the outcome of a golf shot is determined by the impact conditions (the clubhead velocity and clubface orientation at impact and the impact location on the face) but computation of the impact conditions is not simple because the clubhead-ball impact does not allow placement of markers on the clubface in the dynamic trials and the sampling frequency typically used in a golf swing motion capture is not high enough to provide a sufficient time resolution. A swing analysis-specific marker set is required to locate the hand and joint centers. Definition of the local reference frames fixed to the body segments and club (head and shaft) is essential in the angular kinematics. The functional swing plane characterizes the downswing motion of a golfer but computation of its orientation and position requires a special computation algorithm. The purpose of this paper was to identify the unique challenges encountered in golf swing motion analysis, an advanced application of motion analysis, and to present solutions for these challenges.

MARKER SET AND SECONDARY POINTS:

A golf-specific marker set was developed with a total of 55 primary points (tracked markers) including five club markers. In addition, a total of 24 secondary points were defined and computed based on the primary points and the static trials (ball trial, club trial, and setup trial) (Table 1). Appropriate secondary point computation methods (general and joint-specific; wand, mid-point, static point, rigid body, weighted mean, etc.) were selected and configured during the body model setup process. For example, the 'rigid body method' (Schmidt, Disselhorst-Klug, Silny, & Rau, 1999) was used for the arm joints, while the 'Tylkowski-Andriacchi hybrid method' (Bell, Pedersen, & Brand, 1990) was used for the hip joint in the golf-specific body model.

Table 1. Golf Swing Analysis-Specific Marker Set

Group	Markers	Notes
Pelvis (3)	Right & left ASIS and sacrum	ASIS: anterior-superior iliac spine
Leg (8 each)	Greater trochanter (GT), lateral thigh, lateral & medial epicondyles, lateral & medial malleoli, heel, and toe	The GT marker is used in locating the hip joint. The medial markers are used in locating the knee and ankle joints. These markers are removed in the swing trials.
Trunk (4)	Right & left acromial processes, C7, and T12	
Arm (12 each)	Two shoulder markers, three upperarm markers, lateral & medial epicondyles, lateral & medial styloid processes, and three hand markers	The shoulder markers are used in locating the shoulder joints. The medial markers are used in locating the elbow and wrist joints. These are removed in the swing trials.
Head (3)	Three head markers	
Club (5)	Two proximal shaft, two distal shaft, and one clubtoe markers	
Secondary (24)	Ball, eight club markers (four clubface markers, clubhead center, right & left hand centers, and grip end), twelve joints (hips, knees, ankles, shoulders, elbows, and wrists), L4/L5, mid-hand, and hub	The ball and clubface markers and joints require the static trials (ball trial, club trial, and setup trial).

Data collection was conducted in the following steps:
1. A static trial with the ball only was first captured (ball trial). The ball was covered with reflective tape and placed on the tee or at a pre-determined location on the mat.
2. A static trial with the club only was then captured (club trial). The club markers include four shaft markers, four face markers, and one clubtoe marker (Table 1).
3. A static trial with the golfer in the setup position was captured (setup trial). The clubface markers were removed in this trial.
4. Dynamic (swing) trials were captured. Several body markers including the medial markers (Table 1) were removed in these trials as they could interfere with the swing motion.

During data processing, the static trials were registered and the secondary marker locations were computed automatically based on the static trials and the primary markers, including the ball, clubface points, and joint centers.

EVENTS AND IMPACT CONDITIONS:

A total of nine events were defined for the golf swing analysis: Start (ST), Address (BA), Top of Backswing (BT), Upright Club Position during Downswing (DV), Horizontal Club Position during Downswing (DH), Ball Impact (BI), Mid Follow-Through (MF), Start of Impact Zone

(IZ1), and End of Impact Zone (IZ2). The event frames were identified by the automatic event detection function based on a set of standard and user-defined variables.
To compute the impact conditions, the precise locations of the ball and clubhead, and orientation of the clubface at impact must be known and a pre-requisite for this is an accurate determination of the impact instant. When the sampling rate is not sufficiently high, interpolation of the position data is necessary to improve the time resolution. Auto detection of the BI, IZ1, and IZ2 events involved the following steps:
1. The position data were re-sampled (interpolated) at 2,000 Hz. All standard and user-defined kinematic and kinetic variables were re-computed accordingly.
2. The first frame in which the clubface was in contact with the ball was selected as the BI event. Fifteen centimeters before and after the impact position were set to IZ1 and IZ2, respectively. IZ1 and IZ2 define the 'impact zone' for the assessment of the error margins.

IMPACT CONDITIONS AND ERROR MARGINS:

The impact conditions include the clubhead velocity, clubface orientation, and impact location on the clubface at the true impact instant, not BI. The following steps were taken in the computation of the impact conditions:
1. The clubface reference frame was defined based on the clubface markers (Williams & Sih, 2002): X - away from the body; Y - normal to the clubface; Z - upward. The origin of the clubface reference frame was set at the clubhead center, the mean of the four clubface markers (Figure 1).
2. The first frame the Y-coordinate of the ball described in the clubface reference frame became smaller than +2.134 cm (radius of the ball) after the DH event was automatically selected as the BI event. As a result, the true impact instant (when the Y-coordinate of the ball described in the clubface frame becomes equal to the radius of the ball) occurs sometime between the BI frame and one frame before. The true impact instant was determined by the sub-frame data reading function.

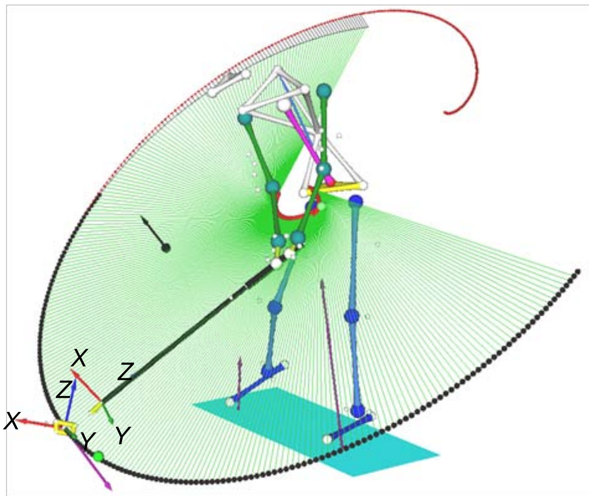

Figure 1. The downswing motion of a golfer. The clubhead velocity vector (pink arrow), clubface frame axes (red, green, and blue arrows), the functional swing plane (black trajectory), rotation centers (red balls), and rotation arms (green lines) are shown.

3. The velocity of the clubhead center was used as the clubhead velocity (Figure 1). The angles formed by the Y-axis unit vector of the clubface frame with respect to the global horizontal and vertical planes toward the hole-cup were used as the clubface orientation parameters. The impact location on the clubface (X- and Z-coordinates) was determined by the ball position described in the clubface reference frame at the true impact instant.
4. The ranges of the clubhead velocity, clubface orientation, and ball position described in the clubface frame within the impact zone (IZ1 to IZ2) were used as the error margins.

FUNCTIONAL SWING PLANE:

The functional swing plane (FSP) characterizes a golfer's downswing motion (Shin, Casebolt, Lambert, Kim, & Kwon, 2008). The FSP is the plane formed by the clubhead trajectory within the delivery zone (DH to MF) (Figure 1). The FSP parameters (planarity, slope, and direction angle) were computed using the 'Plane Analysis' function. A special plane fitting algorithm based on the Newton-Raphson method is used in this function. The locations of the instantaneous rotation centers and radii of the rotation arms were computed (Figure 1).

For the computation of the orientation of the clubshaft to the FSP, the shaft reference frame was defined by the shaft points and the X-axis of the clubface frame: Z - along the shaft toward the grip; Y - perpendicular to the Z-axis of the shaft and the X-axis of the clubface; X - perpendicular to the YZ-plane of the shaft (Figure 1). The angle formed by the X-axis of the shaft with respect to the swing plane was used as the shaft orientation angle.
The mid-hand point and the left shoulder joint (Table 1) were used in finding the motion planes of the hands and left shoulder, respectively. The 'Plane Analysis' function was used for the body motion planes as well. The orientation of the body planes were described with respect to the FSP by defining a series of user-defined variables.

SEGMENT ORIENTATIONS AND JOINT MOTIONS:

Various body orientations and joint motions should be computed in the golf swing analysis based on the local reference frames fixed to the body segments. In a segment at least three fixed points are required to define a reference frame. The golf-specific marker set presented in Table 1 provides a complete array of local reference frames of the golfer's body segments and club.
Reference frames were defined easily along with the relative orientation relationships and rotation sequences (e.g. mediolateral-anteroposterior-longitudinal sequence). In some cases, a segment (e.g. hand-club) was defined multiple times to assign different relative orientation relationships. The forearm segment was divided into ulna and radius and the pronation/supination of the forearm was computed separately from the wrist joint motion. The clubshaft and clubhead segments were also included in the model.
One way to expand the scope of analysis is to define and use various imaginary segments in addition to the real body segments. For instance, the FSP segment was included in the body model as an imaginary segment to utilize the FSP reference frame in the analysis. The shoulder girdle segment (and reference frame) was also defined to analyze the rotational motion of the shoulder about the hub.

SUMMARY:

To meet the unique challenges encountered in various analysis settings, it is important to develop/use a comprehensive motion analysis program characterized by strong mechanical foundation, flexibility, adaptability, and expandability. The program structure must allow the users to easily create/modify body models and the body models should define the mechanical characteristics of the body in detail. A vast array of processing capabilities and functions should be available and configurable by the user to streamline the analysis while eliminating the need for additional user programming. Golf swing analysis is an advanced application of motion analysis in which advanced analysis methods are required to resolve unique challenges encountered. It was demonstrated how the unique challenges could be resolved in golf swing analysis by using a comprehensive motion analysis program, such as Kwon3D.

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REAL-TIME MEASUREMENT USING EMG AND MOTION CAPTURE SYSTEMS

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Real-time feedback to participants of a particular parameter has been a viable tool for learning, training and rehabilitation for many years. The type of feedback that was usually given was in the form of kinematics (e.g. accelerometry) or kinetics (e.g. force). The manner in which the feedback has been given has been altered greatly in recent years by the development of equipment and software that enables the researcher to accomplish the feedback tasks much more easily than previously. In this paper, real-time electromyography (EMG) and real-time kinematics will be presented. The most recent equipment for both real-time EMG and kinematics will be presented along with examples from recent research papers that used such techniques.

KEYWORDS: feedback, electromyography, maximum voluntary contraction, kinematics, functional joint center

INTRODUCTION & OVERVIEW:

For many years, the real-time assessment of movement has been a topic of great interest. Usually real-time feedback was obtained for kinematics using electrogoniometry or accelerometry or for kinetics using force transducers. However, the equipment to easily accomplish real-time assessment for EMG and for motion capture systems has only become readily available quite recently. It should be noted that “real-time” does not necessarily mean at the exact same time as the occurrence of the measurement parameter. Real-time essentially means that there is a very small lag between the occurrence and the presentation of the data. The purpose of this paper is to illustrate the techniques of real-time acquisition and assessement in electromyography (EMG) and kinematics and to present examples of how these techniques can be used in laboratory settings.

REAL-TIME EMG:

For this paper, the EMG equipment that was used included a Delsys Trigno Wireless EMG system with EMGworks 4.0 software sampling at 4000 Hz. It is assumed in this paper that the placement of EMG electrodes is in compliance with usual standards (Basmajian and DeLuca, 1985). In order to perform real-time EMG recording, it may be necessary to normalize the raw EMG data for comparisons between subjects or to track subject progress over time. Since the amplitude of the EMG signal can depend on sensor placement, skin preparation, and other factors, normalizing the data to a maximum voluntary contraction (MVC) can allow the researcher a quantitative method to compare between subjects or test sessions. This can be easily accomplished using real-time methods. MVC can only be elicited with proper subject training when feedback of the EMG level is observed. Without training, a subject may only contract to 60-80% of their true MVC level. Alternatively, a known voluntary contraction could be used for test subjects that may not be able to perform an MVC. In this case, a sub-maximal activity – such as a toe raise or bicep curl with little weight added – can be used to normalize data (see Figure 1).

PRACTICAL APPLICATIONS OF REAL-TIME EMG:

The type of research that real-time EMG can be used for includes:

1) physical rehabilitation - for physical rehabilitation studies, EMG feedback can be used to monitor the use of certain muscles (Horsley et al., 2010). With appropriate coaching and feedback, a subject can be trained to either activate or avoid activating a certain muscle while performing a task (see Figure 2).

2) fine motor control assessment (tracking) - to assess a participant's fine motor control, tracking a constrained trajectory using biofeedback can be used (Prodoehl and Vaillancourt, 2010).. A number of tracking trajectories can be used for different purposes (see Figure 3).

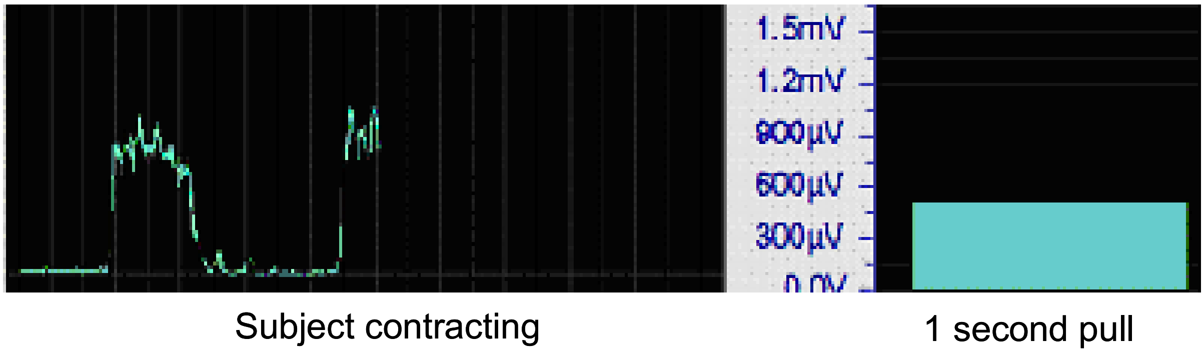

Figure 1. Real-time EMG feedback on the left and a bar graph showing the subject their muscle activation level on the right.

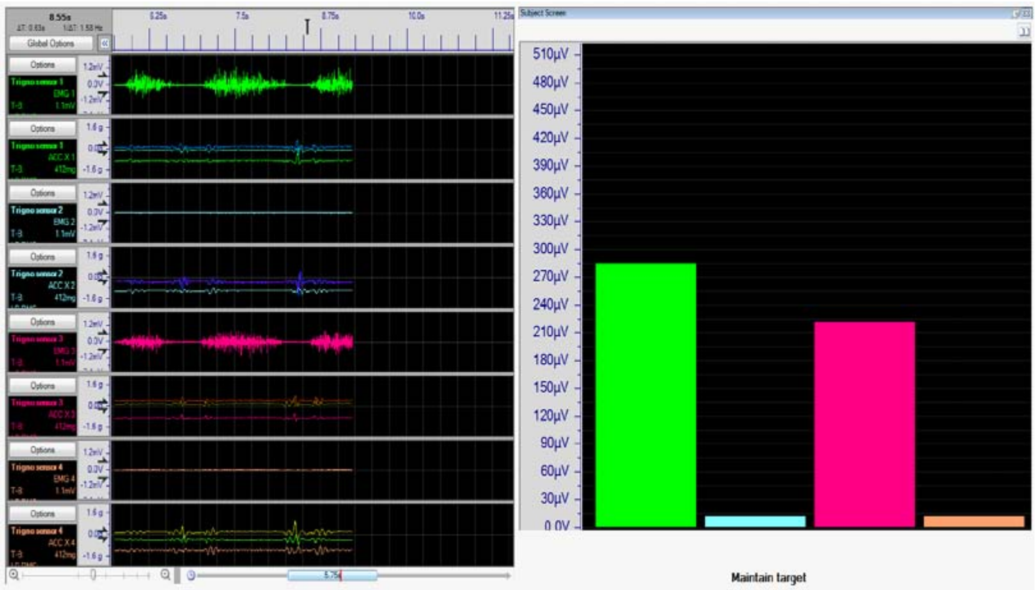

Figure 2. On the right, color-coded bars show the real-time EMG activity to the subject in an easily interpretable fashion—a taller bar means more activity. The subject can use this information to modify their behavior in order to selectively activate or not activate a muscle.

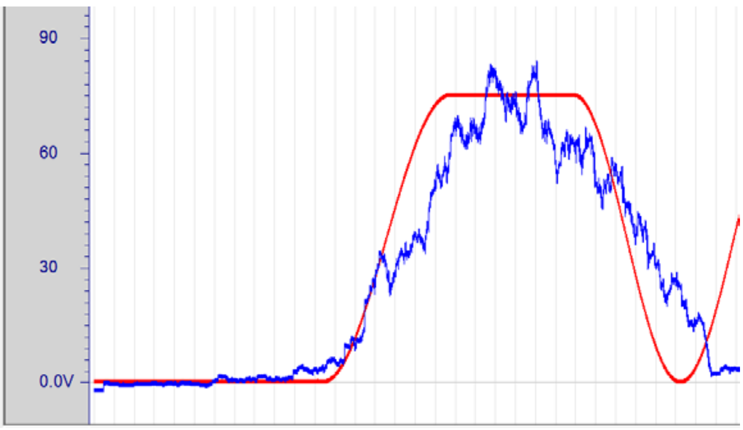

Figure 3. In real-time, the participant is required to activate the muscle is a manner that tracks the target.

REAL-TIME KINEMATICS:

The equipment that was used to present real-time kinematics was an eight camera Qualysis Oquus motion capture system sampling at 250 Hz. The software used for real-time capture is Visual3D (C-motion, Rockville, MD, USA) which can be used in conjunction with all of the major motion capture systems. There are two uses that real-time data capture can serve in a kinematic study. The first of these involves the determination of functional joint centers that will be later used in the analysis. In order to compute a functional joint center, a model must be determined with all of the relevant markers identified. In addition, all of the functional joints must be defined. With the subject standing in the field of view, the markers that define the joint are identified while the participant moves the joint through a range of motion that is functional for that joint. Based on these movements, the instantaneous joint center is defined. There are particular advantages in using a functional determination of the joint center rather than the usual static determination. Most important is that the joints that are determined are functionally relevant and anatomically meaningful. This provides an advantage to further calculations that may be considered more accurate.

After the functional joint centers have been determined, these joint centers are used in the calculation of 3-D joint angles. In the streaming data mode, with the participant performing a particular movement, the markers are automatically identified, the angles calculated and a particular angle may be presented to the participant. The participant may see the angle as a time history or as a bar graph. By viewing this feed-back, the participant may alter or sustain a particular level of joint motion (see Figure 2).

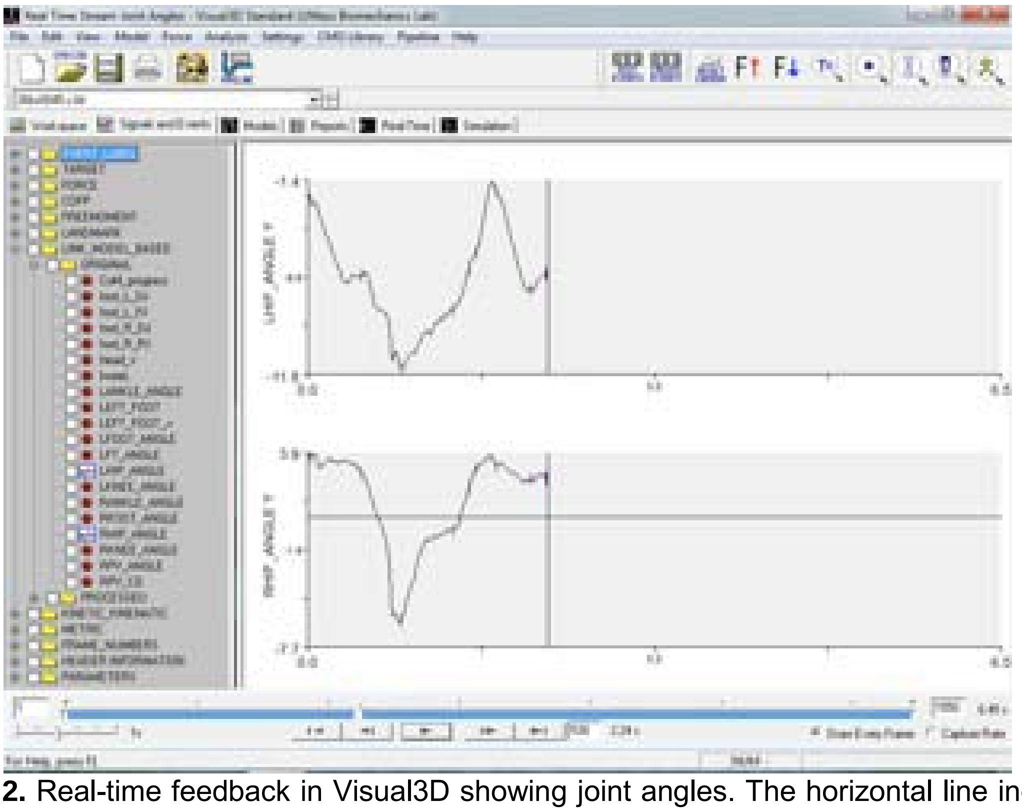

Figure 2. Real-time feedback in Visual3D showing joint angles. The horizontal line indicates the time at which the sample was taken.

PRACTICAL APPLICATIONS OF REAL-TIME KINEMATICS:

Real-time kinematics have been used for several types of feedback studies. In a recent study, Hanlon and Anderson (2005) used real-time bandwidth kinematic feedback in learning a novel cyclical lower extremity movement skill. The task used in this study involved accurately replicating a 4-s movement of the right limb followed by non-feedback trials. These authors reported that steady performance improvements were made with relatively simple feedback.

In a study that incorporated real-time feedback, Page and Hawkins (2003) developed a system in which rowers were trained on an ergometer that gave biomechanical information as feedback. The rowers were presented with a stick-figure representation while they performed on the ergometer. In addition, they were given information on handle position. Re-training movement after an injury has been a key feature of rehabilitation treatment in physical therapy. Willson and Davis (2005) used real-time feedback from a motion capture system on injured runners. These researchers determined a target range of motion for a particular lower extremity joint for each participant. When presented with a real-time bar graph of the target area, the runner attempted to alter their movement pattern within the target range.

CONCLUSIONS:

Real-time data acquisition is not new and the applications of this type of data acquisition have been used previously. However, much easier real-time data acquisition for EMG and kinematics has been made possible by advances in both technology and software. While the research questions have not really changed, these developments have made real-time studies more accessable to researchers and much easier to accomplish.

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FIELD MEASUREMENT OF BIOMECHANICAL PERFORMANCE

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The time course of joint moments and powers during competitive performance are helpful variables for pinpointing the involved major muscle groups, estimating total mechanical energy expenditure and mechanical efficiency, and identifying the magnitude and timing of unnecessary joint power absorption. These measures are also useful for understanding the mechanisms of injury and thus planning and training for their minimisation. Simulation of sporting movements in the laboratory is useful but never reaches the validity of testing in field conditions. However, field instrumentation must meet additional criteria imposed by the environment and the necessity to minimize interference with the athlete's performance. This session will propose some solutions for field testing of athletes.

KEYWORDS: field measurement, motion analysis, technology, inverse dynamics

INTRODUCTION & OVERVIEW:

Background: The combination of biomechanical theory and measurement tools enables biomechanists to describe sporting movements in detail and deduce mechanisms of high performance. Potentially, with adequate measuring equipment, the sport scientist can output time series net joint moment and power data for all important joints of the body. Power generation and absorption can be pinpointed and mechanical efficiency calculated. Once the mechanisms are known, evidence-based training programs can be designed to produce more efficient or effective performance. For example, once it is known that the knee exerts a flexion rather than extension moment in the middle of the rowing drive phase (Smith & Milburn, 1996), the deadlift is exposed as a non-specific strength training modality for the lower limbs. This evidence could only be obtained from inverse dynamics or electromyographic analysis. The issue of what information and how this should be fed back to the coach and athlete will not be dealt with here.
Sports: Track and field events may be performed indoors and more laboratory-oriented measurement systems can be employed. Nevertheless, the SESAME project (http://www.sesame.ucl.ac.uk/) shows how unconventional this can become. Equestrian, cycling and aquatic sports do not lend themselves easily to traditional biomechanical measurement systems and it is in these areas that new technologies will have the highest impact. Further, cycling, rowing, flatwater kayak, canoe, dragon boat feature one or more fixed connections between the athletes body and the vehicle creating the possibility of utilizing these constraints to simplify the measurement system without losing accuracy.
Technology: Developments in microelectromechanical systems, GPS, wireless, microprocessors, batteries, standardisation of communication protocols and software support have opened up a vast array of possibilities for unobtrusive but detailed biomechanical measurement of many sporting movements. Commonly available GPS is now capable of a data rate of 10 Hz and position accuracy of 2.5 m in a 9g 22 x 22mm package.

The aim of this paper is to describe some useful technology for motion analysis in the field and give some examples of its successful application.

METHODS:

Orientation and Position: Inertial measuring units (IMUs) are now low cost (~US$100), accurate (± 1°) and have a high angular velocity rating (1200 °/s). They can be three dimensional and output acceleration, angular velocity, and heading or angular position and linear position. They are small and light and can be conveniently attached to body segments as small as the hand. They can be wireless (appropriate for one or two). From practical

experience, a fifteen segment system should be hard-wired to a central wireless transmitter in a harness or attached to the athlete by a belt.
If the segment is expected to rotate through more than 180º, quaternions could be used rather than Euler angles to avoid the gymbal lock problem. The quaternions can be transformed back to Euler angles once the data has been collected.
Force: Portable force platforms can be constructed from three dimensional transducers sandwiched between two carbon fibre sheets. Our most recent attempt had an accuracy of <1% for all dimensions but required a minimum of four transdcuers. Centre of pressure can also be computed.
The problem is much simpler if one dimension is required. There are many small and light one-dimensional transducers available.
For sweep rowing particularly, three dimensional force sensing is desirable especially for the pin force. However, in flatwater kayak the footbar and seat force are adequately measured in the propulsive direction only. For cycling, there are a number of crank torque transducers available.
Telemetry: On-body and/or on-board telemetry can be combined to provide a minimally wired system. Bluetooth can be used for short range and Wifi 802.11b/g/n for distances up to a km or two (Figure 1.).

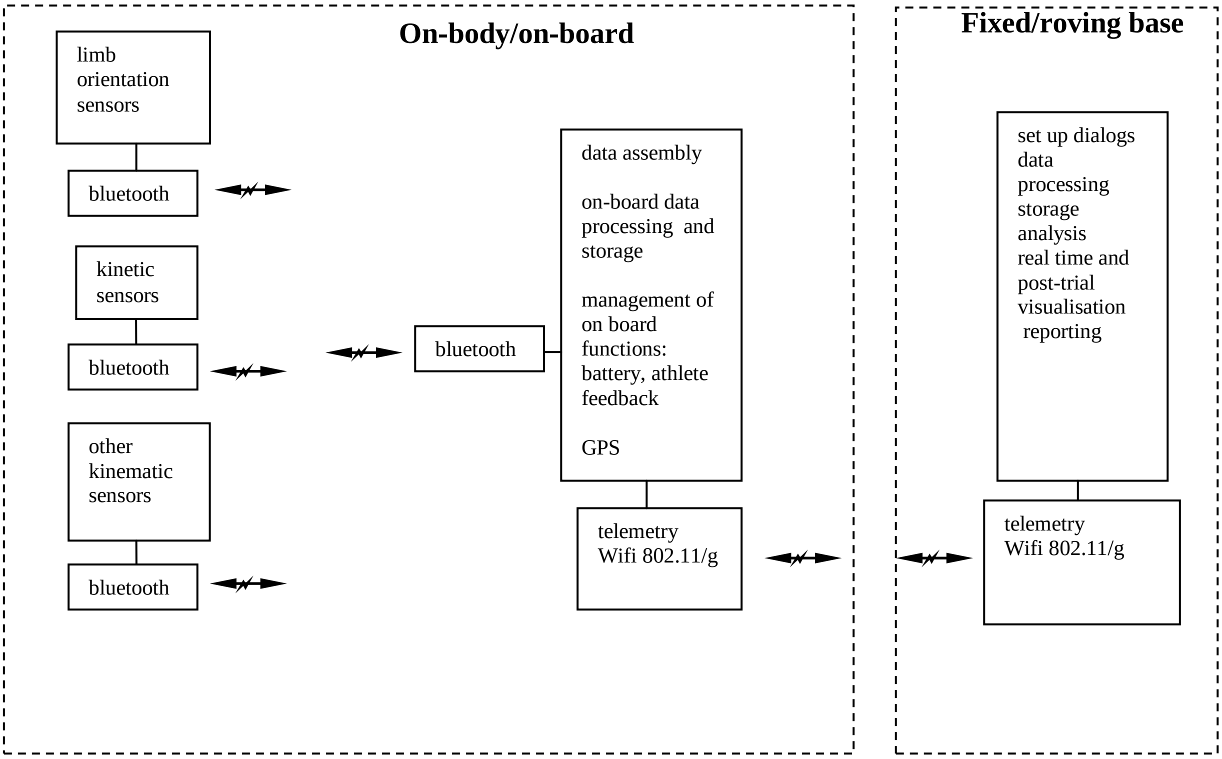

Figure 1. Configuration of wireless motion capture and analysis system.

RESULTS:

Dragon Boat: Elite dragon boat paddlers have more efficient paddle strokes than sub-elite paddlers (Figure 2.) (Ho et al., 2009). The data for this result was obtained using calibrated strain gauge transducers mounted on the inside of the paddle shaft. Equation 1 was used to calculate the efficiency.

$$Efficiency = \frac{\Sigma(\Delta \mathrm{Energy}_{\mathrm{propulsive}})}{\Sigma(\Delta \mathrm{Energy}_{\mathrm{total}})} = \frac{\Sigma(\mathrm{Force}^{\mathrm{blade}}_{\mathrm{propulsive}} \times \Delta \mathrm{Distance}^{\mathrm{blade}}_{\mathrm{propulsive}})}{\Sigma(\mathrm{Force}^{\mathrm{blade}}_{\mathrm{normall}} \times \Delta \mathrm{Distance}^{\mathrm{blade}}_{\mathrm{normal}})} \qquad \text{(equation 1)}$$

The paddle strain gauge outputs were conditioned electronically and input to an analog to digital module before sending to a custom telemetry device operating in the 900 MHz band.

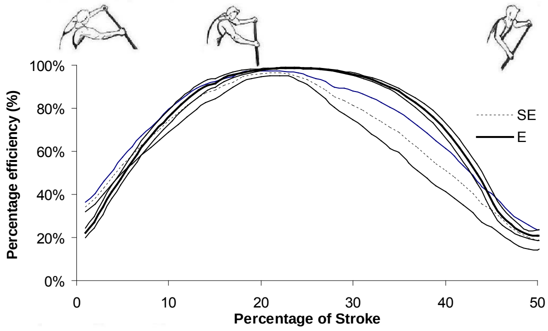

Figure 1. Average (±95% confidence interval) dragon boat paddling efficiency of sub-elite (SE) and elite (E) groups during the drive phase (normalised to one stroke).

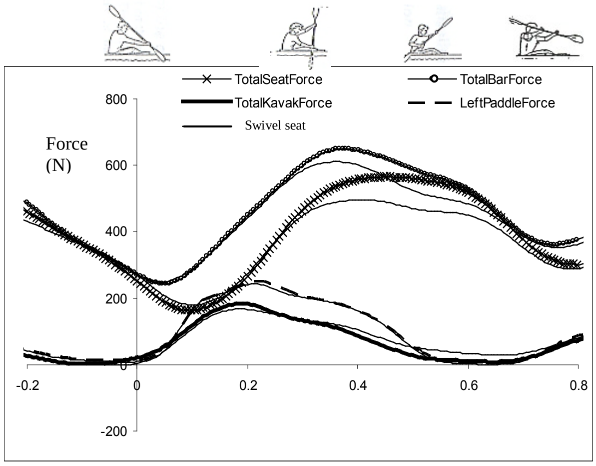

Figure 2. Flat water Kayak paddling. One stroke with the left paddle in the fixed seat compared with the swivel seat (thin solid lines).

Flatwater Kayak: The net propulsive force on a single kayak was measured on-water while the paddler was using a fixed seat then a swivel seat (Figure 2.). The footbar force, seat force and the normal paddle force were measured in one dimension. The average force on the footbar and seat is less for the swivel seat. The knee extension muscle force would be less on average and leads to a greater capacity for power output for a given metabolic cost (Michael et al., 2010).

Rowing: The transverse pin forces in a pair boat cause an unbalanced moment about a vertical axis which potentially would cause boat yaw. The on-water instrumentation system reveals that pair rowers minimize boat yaw by compensating for the unbalanced horizontal plane moment by differentially manipulating the magnitude of the propulsive force on the pins (Figure 4) (Smith and Loschner, 2002). Three dimensional piezoelectric force transducers were used at the pin. The outputs were electronically conditioned and input to an analog to digital module that provided serial output to a custom telemetry unit. Total power output to the boat can be calculated (Figure 5).

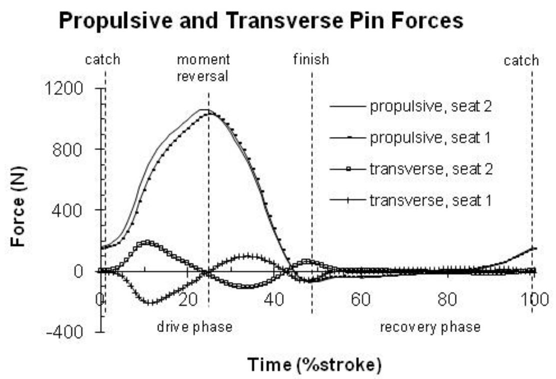

Figure 4. Propulsive and transverse pin forces for elite rowers in a pair boat. The unbalanced moment caused by the transverse forces is compensated for by the magnitude and timing of the propulsive force.

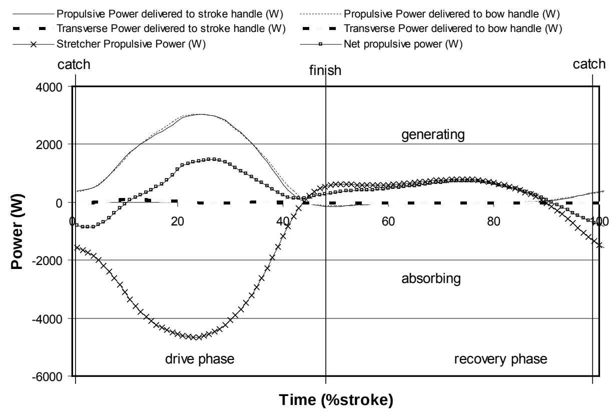

Figure 5 Ensemble average pair power output time series at 30 strokes per minute.

PRACTICAL APPLICATIONS:

The information gathered by the described systems can be used for crew selection as part of a performance tracking system or individually tailored for individual athletes and their coaches. Particular aspects of the athletes technique can be targeted, a plan made for the changes required and further measurements conducted to see if those changes have been learned.

More research work needs to be carried out into the selection of information that is used for feedback to the coach and athlete and into the different methods of providing the feedback. The systems described in this paper have the potential to provide real time feedback of raw and calculated data for both athlete and coach. For example a single number could be calculated that is an estimate of the efficiency of the rowing stroke at the completion of each stroke. Alternatively the time series of force(s) or angle(s) could be presented in real time. No research has been conducted yet which helps with this decision-making process.

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Validation of an Outdoor-based Passive Optoelectric Motion Capture System

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The purpose of this study was to validate the quality of data captured outdoors in full sunlight using a passive optoelectric camera system. A golf swing analysis was performed outdoors and indoors using the same system; the outdoor collection was performed in full sunlight. Golf club rotation (deg) and angular velocity (deg·s^{-1}) data were calculated about the X, Y, and Z axes of the golf club for a single male subject. Outdoor and indoor angle and angular velocity data were similar about each of the three primary axes of the club, with r values≥ 0.970. The highest correlation values were found to exist among the angle data. This study demonstrated that data quality captured with an outdoor system is comparable in quality to data captured indoors.

Introduction: Many motion capture systems measure gross human movement, including photogrammetric (Bergemann, 1974; Marzan and Karara, 1975; Miller and Petak, 1973; Shapiro, 1978), optoelectric (Greaves, 1983; Scheirman, 1992), magnetic field (Luo, Niebur, and An, 1996), accelerometric (Frisch, 1989; Breniere and Dietrich, 1992; Lafortune, Henning and Valiant, 1995) or goniometric (Chao, Laughman, Schneider and Stauffer, 1983; Strathy, Chao and Laughman, 1983). Each of these various methods has unique benefits and deficiencies. However, within the last decade, optoelectric methods have become the preferred tool to measure human movement. Exceptional measurement accuracy and a wide range of acceptable image capture equipment are largely responsible for this preference. Currently, optoelectric technologies image the movement with specialized high resolution chips, and then use processors within the camera to identify retro-reflective markers and compute their positions in the image. These markers can be discs, hemispheres or spheres that are covered with retro-reflective coatings. Infrared LEDs ringing the cameras cause these markers to contrast with the background, thus permitting the system to detect their positions in real-time. Typical applications encircle four or more cameras around the space where the action is to take place. From one to hundreds of retro-reflective markers are attached onto the subject at the locations of interest for tracking. Two-dimensional coordinates are passed to the computer, usually via gigabit Ethernet, where software on the computer determines in real-time the three-dimensional coordinates of the markers from data seen by two or more cameras. This method has been well accepted in the biomechanics field because it does not require expensive, calibrated equipment, allows flexibility of camera locations and has proven to be adequately accurate.

Once cameras are calibrated, the system is ready to capture data from markers placed on the subject. Sophisticated template matching routines are then used to create frame-to-frame marker paths over time and resolve any possible occlusions. Identifying templates are defined allowing subjects to be identified in real-time. The real-time identification allows kinematic and kinetic data to be calculated, typically with less than 10ms of latency, giving coaches and trainers instant feedback on performance. These calculations can also be used with sophisticated biofeedback algorithms that give coaches the ability for behavioral modification training.

As robust and as widely used as optoelectric systems are, they all have had one critical limitation: systems that track strictly passive retro-reflective markers, and most that track active LED markers, could not be used outdoors in full or partial sunlight. The light from the camera ringlights has been insufficient for passive optoelectric systems to differentiate the marker from the ambient background illumination. In addition, cameras have not had the ability to process the large amount of data resulting from the sunlight and typically get backed up with data before being sent to the analysis computer. It is for these reasons that passive optoelectric systems have been relegated to indoor use only or to outdoors at night. If outdoor capture in full sunlight was desired, video-based systems (e.g., Innovision Systems, Dartfish) were required. Unfortunately, these video-based systems can have accuracies around 1mm (Richards, 1999), and accuracy tends to diminish rapidly if the system requires manual digitization of the markers. Needless to say, passive optoelectric capture in full sunlight has been a much desired and, in some cases, a needed feature of 3D sports motion capture, particularly when sub-millimeter accuracy is required.

The purpose of this study is to validate the quality of data captured outdoors in full sunlight using a passive optoelectric camera system. To accomplish this task, an analysis of golf club swing kinematics was performed using data collected both indoors (out of the sun) and outdoors in full sunlight.

Methods: Kinematic data were recorded using eight high-speed optoelectric cameras (Raptor-E, Motion Analysis Corporation, Santa Rosa, CA, U.S.A.; 200Hz) and collected using Cortex 2 software (Motion Analysis Corporation). Previous research has shown sampling golf swing data at 180Hz is sufficient for data capture (Mitchell et al., 2003; Nesbit, 2005), but to allow synchronization with other video devices, 200Hz was chosen. Cameras were positioned around the volume to ensure that each marker was visible to at least two cameras at all times throughout the data collection volume. Data were further reduced in post processing using Cortex 2. Golf swing marker data were smoothed with a zero-phase, fourth-order Butterworth filter with a cutoff frequency of 12Hz (Mitchell et al., 2003). The golf swing cycle was defined using events created in Cortex 2. All data were collected from a single male subject (A: 33 y/o; H: 1.73m; M: 81.6kg).

A marker set consisting of 37 strategically placed markers were applied to the subject and golf club. Markers were placed at appropriate locations to define joint centers and track segmental motion. Markers were adhered to the subject using Velcro patches placed over the clothing and directly to the club using adhesive tape. Kinematic data from the markers placed on the subject were used only as a reference and will not be reported in this study.

Four markers were placed on the golf club at the following locations: 50% and 75% of the distance from the top of the handle to the end of the shaft, heel, and toe. The club was modeled as a rigid object to simply data analysis, using an XYZ convention similar to that described by Nesbit (2005). The club X-axis is defined as the axis about which the swinging action is performed with positive pointing forward; the Y-axis represents the pitch of the club with positive to the club's right; the Z-axis represents the rotation about the long axis of the club with positive pointing down the shaft towards the club head. The global coordinate system was defined as X pointing to the subject's rear, Y pointing vertically, and Z pointing towards the direction of the target.

A foam golf ball was placed on a synthetic grass mat and the subject was asked to swing a golf club of his choosing (8-iron). The subject was allowed as many practice swings as necessary and data were collected until three successful trials were completed. A hitting net was placed 2.4m in front of the subject and a successful trial was defined as the subject successfully hitting the ball into a 0.6m x 0.5m square on the net. The square location and size was created based on the resulting ball locations from the practice swing hits.

Golf club rotation (deg) and angular velocity ($deg \cdot s^{-1}$) data were calculated in the KinTools RT software (Motion Analysis Corporation) using the XYZ rotation convention (Nesbit, 2005). The swing phase was defined as the top of the backswing (minimum club head speed) to ball impact (maximum velocity along the global Z axis). Club follow-through was ignored for the purposes of this study. Club angle and angular velocity data were exported as ASCII files from KinTools RT and additional processing was performed in MATLAB (MathWorks, Inc., Natick, MA, U.S.A.).

Kinematic data were time normalized and averaged. Data similarity was compared across the indoor and outdoor golf swing angle and angular velocity data using the Pearson coefficient (r).

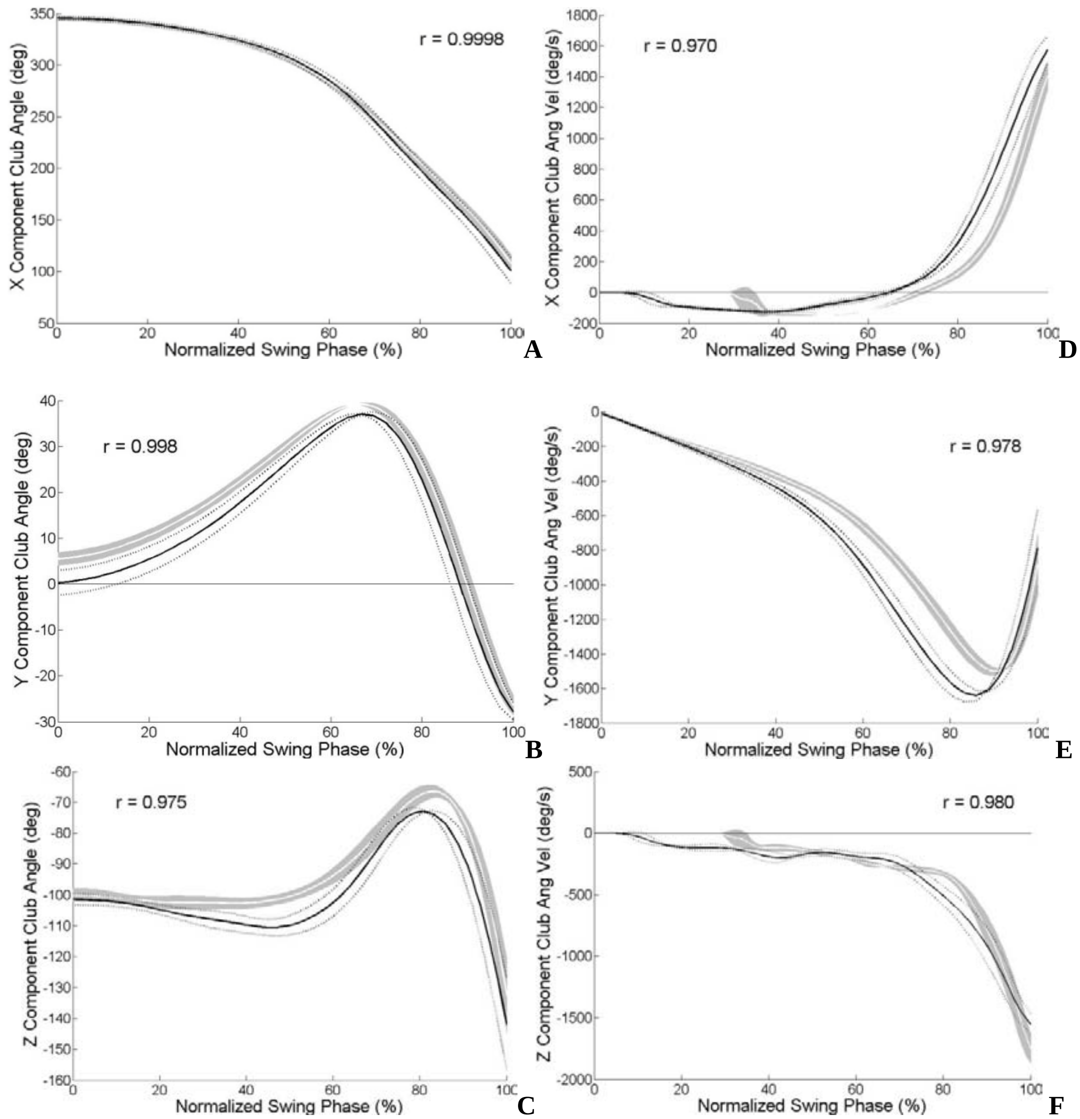

Figure 1. Golf Club Angle (A-C) and Angular Velocity (D-F) over the Normalized Swing Phase (%). The Swing Phase is defined as the maximum take-away position to ball impact. Mean Indoor Golf Club data (white line) ± 1 standard deviation (grey area) shown alongside Mean Outdoor Golf Club data (solid black line) ± 1 standard deviation (dotted black lines). Pearson correlation-coefficients (r) are shown. The X component (A,D) represents the swinging action of the club; the Y component (B,E) represents the pitch motion of the club; the Z component (C,F) represents the rotation about the long axis of the club.

Results: The golf club angle and angular velocity data overall were similar for the motions about each of the three axes. The indoor and outdoor data all had r values ≥ 0.970 (Figure 1).

Discussion: Overall, the outdoor club angle and angular velocity data showed a high correlation with the indoor data (r≥ 0.970). The lowest correlation values, in general, were found to exist between the indoor and outdoor angular velocity data. This is not surprising since environmental factors such as heat and glare, which were not present with indoor captures, were mentioned to be possible issues by the subject, potentially affecting the rate of club swing. The X and Z components of the club angular velocity agree with previously published data, but the Y component only appears to follow the general trend (Nesbit, 2005).

Data captured outdoors are subject to environmental factors which can affect system accuracy. In addition, various cognitive factors (e.g., feel, comfort) can affect how a subject behaves in various environments, thereby altering how a subject would perform an activity.

This study shows that not only is it possible for passive-based optoelectric motion capture systems to collect data outdoors in full sunlight, it is possible for those data to have comparable quality to data collected indoors.

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ISBS 2010

Applied Session

Feedback to Athletes

Biomechanics Feedback to Olympic Athletes – Two Feedback Models

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KEY WORDS – Biomechanics, Kayaking, Hockey, Skill Acquisition

INTRODUCTION/OVERVIEW: One of the challenges for biomechanics, as a discipline in general, has been to provide meaningful and timely feedback to athletes. This is particularly the case where the biomechanist is working with athletes frequently over a period of time. A number of different approaches in providing feedback to athletes can be made depending upon the question asked, testing setting, and the data collected. These issues need to be balanced when dealing with Olympic athletes.

The Australian Insititute of Sport biomechanics programme has been working with Olympic athletes for over 25 years now. Essential in this programme is the collection of high quality biomechanical data and importantly, the feedback of this information to the athletes. This provision of feedback has seen a number of changes during this 25 years. Initially, feedback models followed a service based approach typified by frequent testing and quick feedback. More recently the environment has changed where a research/project based model is increasingly used in applications where this model is considered more optimal. Another important development has been the addition of Skill Acquisition specialists to the AIS in 2002. There is an ideal symbiosis between biomechanics and skill acquisition, where the main aim is to improve athletes' technique and performance. Meaningful feedback and application of the results and findings of testing is essential.

Service Based Model

A service based model is where tests and analysis are conducted on a regular basis with relative ease and minimal disruption to the training and/or competition program (Baker, 1998; Sperlich & Baker, 2000). Results are then provided to the athletes and coaches in a short space of time so that any findings or recommendations are able to be implemented. Research and/or data mining underpins the variables used as the prime analyses foci.

Prior to the commencement of a scholarship with the AIS Sprint Kayaking program athletes complete a VARK (Fleming et al., 2005) questionnaire to determine their preferred learning style. Where possible the feedback provided to the athletes is then catered around their learning styles.

In light of previous work for AIS sprint kayaking (Baker et al., 1997), the emphasis of the servicing program was on-water force testing. Using a force system developed by the AIS (Hunter & Davis, 2007), it has been possible to test paddlers' on-water and then shortly after, show them results of their force related, and boat movement, parameters. Two of the key factors in sprint kayaking performance are the peak force applied to a paddle during a stroke, and the related impulse. In addition the shape of the force-time curve is important (Baker et al., 1999). With the data and video combined, and the respective coach present, feedback is provided in a group setting where the preferred learning style of the athlete is catered for in the feedback.

In addition to the on-water testing, servicing includes race planning and race analysis. This work is important to provide feedback on athletes' race performance and to improve aspects of the athletes' race plan. The use of GPS and MEMS based sensors are used for this testing.

Project Based Model

A project based model is set up to answer a specific research question. For the Australian Hockey team it was to improve goal keepers' ability to defend goals in short corners, specifically for a drag flick. This project was used as preparation for the 2004 Olympic Games.

Task complexity is a major issue for goalkeepers in a short corner. Apart from the shot itself there are a number of confounding factors to deal with such, as variation plays, and attackers and defenders being actively involved. However the over-riding issue, as verbalised by the goalkeepers and coaches, was the ability to determine drag flick direction.

Based on this task analysis, a three step project was designed:

1. A 3D motion analysis of the drag flick to determine the key kinematic factors.
2. Temporal occlusion testing to help ascertain when goalkeepers start to pick up visual cues from the drag flickers. The temporal occlusion epochs were based on the biomechanical analysis so as specific facets of the movement, which determined direction, were contained within the presented visual phases. Within the temporal occlusion testing a moving-window design was also included to determine whether direction prediction is driven by the content of the visual information presented rather than available processing time.
3. A video-based temporal occlusion training programme to support the on-field training the goalkeepers perform with the team and the goalkeeping coach. The temporal occlusion testing determined information pick-up occurred very late in the drag flick movement. From this result, combined with biomechanical findings, the occlusion training was orientated towards late in the movement.

The outcomes of the project were:

1. The key kinematic parameters of the drag flick and when they occurred in the movement was identified.
2. From a visual-perspective aspect when in the movement goalkeepers started to pick-up information was determined. Furthermore, the occlusion testing determined that kinematic content of the occlusion condition rather than processing time is responsible for the anticipatory skill of goalkeepers (Baker et al., 2010).
3. The video-based training programme increased prediction rates of drag flick direction.

CONCLUSION: Two common methods of feedback used at the AIS have been presented with both achieving their objectives. The Servicing Based Model is used when there is a need for on-going monitoring of athletes' performance. The Project Based Model is typically used when there is a need to answer a specific question, and is often used with sports where there isn't an on-going need for biomechanical analysis. Each feedback method also illustrates the benefit of combining biomechanics and skill acquisition in the elite sport environment. As an addition, task representative design of testing has been a constant issue for biomechanists. Motor learning theorists, principally those from the dynamical systems approach (Arujao et al., 2007), would argue the ecological validity of testing is omnipotent. As biomechanics advances these theories will impact more on our feedback methods, and will need to be addressed.

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AUGMENTED FEEDBACK – THE TRIPTYCH CONUNDRUM

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KEY WORDS – motor skills, real-time augmented feedback, feedback technologies

INTRODUCTION - The area of artificially augmented feedback of biomechanical parameters has received increased attention over the past decade; the advancement in miniature sensors and increased computer speeds has made real-time artificially augmented feedback of data (rtAF) commonplace in sport. The conundrum which exists in biomechanics, having easy access to technologies enabling rtAF, is ensuring an effective triptych – (1) the feedback must be accurate and relevant, (2) the feedback must be timely and delivered correctly, and (3) the feedback must be decipherable by the athlete. Research over the past decade in our group at the University of Limerick has attempted to shed light on this conundrum.

The first part of the triptych must be related to the type of feedback technology – our research has reported that elite coaches, when presented with either still or video images, are unable to correctly identify simple angle measurements accurately (O'Halloran & Anderson, 2005). This finding sets the scene for non-subjective feedback protocols to be considered. The traditional approach of oral feedback after visual observation must, therefore, be augmented by artificial sources of data. The development of sensors such as MEMS based accelerometers, gyroscopes, GPS etc. enables the biomechanist to design sensors which can provide accurate data to the coach/athlete/scientist combination. The decision on whether this is relevant data is then required; traditionally this involved subjective discussion between the coach, athlete, and scientist. Our current research (see Tucker et al., 2010) is attempting to answer this question with mathematical modelling related to golf; if we can alter the club head velocity (the outcome measure) by altering, for example, the variability at the right knee joint, it illustrates that this parameter has a direct influence on outcome; thus this may be worthy to be considered as a data source in the feedback protocol. Part one of the triptych conundrum is complete.

The second part of the triptych requires communication between the sports psychologist (or motor skill specialists) and biomechanists. The variety of ways this feedback data can be presented to an athlete is limitless and only governed by the imagination of the software programmer. Whether this feedback is to be delivered aurally or visually, concurrent or post-performance, blocked or random are questions that require consultation. The field of motor skill learning and development contains the answers, or at least the route to the answer, to most of these questions. However, the lab based experiment is still required; there is not a single scenario that suits all applications. We have completed work using aural feedback in gait (Hanlon & Anderson, 2005), visual feedback in rowing (Anderson et al, 2005), concurrent feedback in golf (Fitzpatrick & Anderson, 2007), post-performance feedback in swimmers (Meehan & Anderson, 2004) and numerous other combinations which have all resulted in positive results for feedback. The vital link between these research studies was the consultation with the motor skills specialists (see Buttfield et al, 2009 for further discussion). Part two of the triptych conundrum is now complete.

The third part of the triptych is often overlooked by the scientists – the ability of the athlete to decipher and process the information provided to them via the feedback loop. If the feedback is too complex and delivered concurrently the athlete has a propensity to switch off. There is anecdotal evidence from our work that, if the feedback is delivered visually, the athlete just closes their eyes and removes the feedback. Preliminary research examined this question from two perspectives. Firstly, during a maximum effort task (2000m row) athletes were asked to respond to either a verbal instruction, a visual distraction, or a combined task (O'Leary & Anderson, 2002). The results indicated that overall performance did not change when the athlete carried out these tasks – the implication of this is that the athlete has a reserve of processing power even when participating in a maximal effort highly complex motor skill; therefore feedback of information that requires some processing is acceptable. Secondly, it seems very beneficial for the athlete to understand the direct link between their actions and the resultant outcomes. During a simulated line-out throwing task in rugby one group was provided with two 1 hour sessions on the physics of the ball flight, impact of ball rotation, projectile motion etc. and outperformed the control group in the post test (Anderson et al., 2005). Resultantly, we must enable the athlete with the skills to interpret and decipher the information being provided to them within the feedback protocol. Hence, part three of the triptych conundrum is now complete.

CONCLUSION - Feedback of biomechanical data in sport can be very successful, and lead to performance gains in both outcome measures and movement characteristics; both essential for the advancement of sport. However, to gain this success consideration must be given to all parts of the triptych. Each part must support the other, if one fails the whole triptych fails – and the conundrum will then remain unanswered. The presentation will discuss this conundrum in more detail and outline the process we have gone through at the University of Limerick to ensure our triptych is in place and functions well.

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BIOMECHANICS FEEDBACK IN SWIMMING

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KEY WORDS: swimming, biomechanics, feedback.

ABSTRACT
Feedback in swimming may include measures of physiological and technique variables related to performance. Among highly trained swimmers, who are approaching their physiological limits, fine-tuning technique is essential to realising performance potential. Feedback on swimming technique ranges from analysis of performance in races to feedback from in-depth three-dimensional (3D) quantitative analysis of technique. The purpose of this paper is to describe and evaluate examples of feedback provided to swimmers and coaches by analysts of the Centre for Aquatics Research and Education (CARE) of Edinburgh University. These include the following:

1. Immediate replay on poolside of above and below water video recordings of mid-pool swimming, starts, and turns, with spontaneous qualitative analysis and interaction among biomechanists, swimmers, and coaches.
2. Rapid quantitative analysis of gliding ability and spontaneous qualitative assessment of gliding performance in relation to postures.
3. Simple 2D quantitative analysis supplementing qualitative video analysis presented to swimmers, coaches, and support personal such as physiotherapists and strength and conditioning specialists. The analysis includes quantification of stroke length, stroke frequency, swimming speed, mid-pool, start, and turn times, and postural assessment with graphical enhancements. A team approach to developing training and intervention strategies is applied.
4. Quantitative analysis of gliding ability and gliding postures and assessment of accuracy of timing the post-glide actions.
5. 3D analysis of swimming technique and postures to identify technique flaws and asymmetries that may be corrected through training interventions. These data are combined with land-based posture, flexibility, and strength assessments to determine appropriate dry land training and in-water intervention programs.

These will be discussed in the light of emerging technologies and analysis systems that improve the turnaround time for feedback and facilitate more efficient, affordable, ‘swimmer/coach friendly’, and convenient evaluation of swimming technique.

ISBS 2010

Applied Session

Writing and Reviewing

HOW TO WRITE MANUSCRIPTS AND RESPOND TO REVIEWERS' COMMENTS FOR 'SPORTS BIOMECHANICS'

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The purpose of this paper is to outline the fundamental writing principles for the manuscripts to be submitted to Sports Biomechanics and how to respond to the reviewers' comments after a review. General and section-specific writing guidelines are presented along with the recommendations for the preparation of the point-by-point responses to the reviewers' comments.

KEY WORDS: publication, writing principles, manuscript format, peer-review, author instructions.

INTRODUCTION:

The ultimate end of a research study is dissemination of the findings through publications. The peer-review process is a safeguard of the quality of the scholarship in the publications and a manuscript must survive the peer-review process to make a publication. It is of crucial importance for the authors to develop a high-quality manuscript to pass the first round of the review without being rejected and to revise the manuscript successfully in response to the reviewers' comments after a review.
The average acceptance ratio of the manuscripts submitted to 'Sports Biomechanics' (SB) for the last 12 months was approximately 27%, which means about 3/4 of the manuscripts submitted were rejected at various stages of the peer-review process. Reasons for rejection include unsuitable topic, poor writing (paper structure, expressions, and grammar), inadequate sample size, poor scholarship (study purpose/questions, justification of the study, and research design), etc., and the majority of the manuscripts were rejected after the first round of the review due to poor scholarship in the manuscript. The purpose of this paper is to outline the general manuscript writing principles for the manuscripts to be submitted to SB and recommendations for the preparation of the responses to the reviewers' comments.

STANDARD MANUSCRIPT FORMAT:

'Original Research' and 'Teaching Biomechanics' manuscripts should be organized in the standard format: 'Abstract', 'Introduction', 'Methods', 'Results', 'Discussion and Implications', 'Conclusion', 'References', 'Tables', 'List of the Figure Captions', and 'Figures'. 'New Method and Theoretical Perspective' manuscripts are also expected to follow this format in general with minor deviations. A 'Review' or 'Letter to the Editor' may deviate significantly from the standard format.

GENERAL WRITING PRINCIPLES:

1. Read thoroughly and comply with the author instructions. The current author instructions can be found at the SB Manuscript Central site (http://mc.manuscriptcentral.com/rspb).
2. The manuscript must be tightly organized around the study purpose and questions. For example, do not introduce any surprises in the 'Results' section that were not mentioned in the 'Methods' section. Do not include any items that are not related to the study purpose/questions. Also place items in the right sections. For example, do not place any method items in the 'Results' section.
3. The manuscript should be concise but sufficiently in detail. Avoid lengthy descriptions of the items presented in detail in other publications. While keeping the manuscript concise, provide sufficient details of the aspects unique to the study so that others can replicate the study based on the information provided. The manuscript should be reasonably self-contained.

4. The writing should be succinct. Avoid wordy sentences and paragraphs. Hit the main points with clarity. Avoid lengthy paragraphs but keep the paragraphs in reasonable lengths with a key theme in each paragraph. Long paragraphs are at best confusing and distracting.
5. Use figures, tables, and equations strategically as they can enhance the manuscript's readability substantially. A figure can be as effective as hundreds of words. A lengthy list of descriptions/items may be replaced with a table. Equations can clearly show the definitions of the key parameters.
6. The manuscript should be reader-friendly. Avoid any attempts that may hurt the manuscript's readability such as an excessive use of non-standard abbreviations. While avoiding colloquial expressions not suitable in scientific writing, make sure to use plain English as much as possible. It is also important to remember that not everyone is an expert in the content area and technical terms need to be defined and explained properly for non-specialist readers.
7. The manuscript should be free of spelling/grammatical errors as much as possible. Run the spelling/grammar checker before submitting the manuscript. For non-English speakers, it is essential to involve someone fluent in English in the writing process. A manuscript can be rejected immediately by the editor on the sole basis of poor English/writing.

SECTION-SPECIFIC WRITING PRINCIPLES:

Title: Use an informative and specific title which reflects the research purpose/design of the study. A good title allows readers to predict the contents of the paper. Avoid using too abstract or broad a title.
Abstract: Summarize the main findings of the study and conclude with clear statements off the research questions of the study. Refrain from including explanatory statements in the abstract. Include the purpose, key methods, main results (including values and statistical results), and conclusions in a single paragraph. Exclude any unnecessary details.
Introduction: Keep in mind that the main purpose of the 'Introduction' section is to justify and present the study purpose. Develop a novel study purpose and a set of meaningful and specific study questions. Along with the justification of the study purpose, an underlying theoretical framework should be included. Plain descriptive studies with no specific research focus or theoretical framework won't be accepted. Include study questions that can be answered by the theoretical framework and study design, not by subjective observations/speculations.
The 'Introduction' section is not the place for an extensive literature review. Present a big-picture literature review to justify the study purpose, instead of detailed reports on individual studies. Include the justification of the study design (factors) used in the study as well. It is recommended to organize the 'Introduction' section in 4-6 paragraphs in about two double-spaced pages.
Methods: Build the 'Methods' section tightly around the study questions so that it contains only the method items that are essential for answering the study questions. Avoid lengthy descriptions of the items that can be found in detail in published papers. While using the references strategically makes the manuscript concise, it is important to provide reasonable details to keep the manuscript sufficiently self-contained. Include sufficient details of the method items unique to the study so that others can successfully replicate the study with the information provided.
The 'Methods' section typically contains items such as participants, trial conditions, experimental setup, data collection procedures, data processing and variable computation, data analysis, and statistical analysis:
1. Include sufficient participant information (such as physique, age, and skill level) as interpretation of the results can be influenced by the participant characteristics. It is essential to use an appropriate sample size. Studies using a small sample size without a strong justification can be rejected immediately by the editor with no further consideration.
2. Present clearly the details of the trials collected, such as the movements studied, repetitions, and specific trial conditions.

3. Include concise but sufficiently in-depth descriptions of the data collection and processing methods.
4. Pay attention to the inflation of the experiment-wise Type I error when multiple comparisons are conducted (Knudson, 2009). Descriptive studies with no specific research question tend to include a large number of dependent variables and neglect the experiment-wise error while making an equally large number of decisions. A specific study focus helps authors to reduce the number of dependent variables.
5. Carefully select meaningful dependent variables that are directly associated with the research questions (hypotheses) and pay attention to the inter-dependencies among the selected variables.

Results: Keep the 'Results' section as concise as possible with only the result items that are directly associated with the study questions. This section is not the place for interpretation of the results. Do not include any method or discussion items in this section.

Use tables and figures strategically. With most numeric data (descriptive statistics) placed in the tables and figures, authors can focus on the main factor effects and the text becomes less distracting. Refer to the tables and figures parenthetically wherever possible. Avoid presenting redundant numeric data in the text that are presented in the tables and figures.

Do not distinguish 'p < 0.01' from 'p < 0.05'. Stick to the critical significance level presented in the 'Methods' section only. Avoid using expressions such as 'almost significant' or 'strongly significant'. Refrain from using 'p < 0.05' repeatedly in the text. Provide the effect sizes (Knudson, 2009).

Use SI units and do not give results to too many significant figures; consider how accurately you can measure or calculate a variable and determine the appropriate number of decimal places accordingly.

Discussion and Implications: This section also should evolve from the study questions. Do not include discussion items that are not directly related to the study questions. This section is not the place for any ungrounded subjective speculations/observations or generic explanatory statements. Items typically found in this section include in-depth interpretations of the key results of the study, comparison with the previously published data, implications of the results, validation of the methods (if placing it in the 'Methods' section is not possible), limitations of the study, and suggestions for future studies.

Separate this section from the 'Results' section so that practitioners can skip the 'Results' section but understand the findings of the study, and their implications for sport performance or injury prevention, from this section alone.

Conclusion: Summarize the main scientific findings in the context of the study's purpose and questions. The conclusions (answers to the study questions) should be provided by the research design, not by subjective observations and speculations.

References: Ensure that the paper contains adequate referencing. Only the references used in the paper body ('Abstract' to 'Conclusion') should be included in the 'References' section. No reference used in paper body should be omitted. Check reference details, ensuring that all references given in the main body and in the reference list correspond in terms of the author(s) and publication year. Follow the SB citation and reference formats outlined in the author instructions (http://mc.manuscriptcentral.com/rspb).

Tables: Place the tables after the 'References' section. All tables should be referred to in the paper body. Keep the table caption as concise as possible but sufficiently meaningful. Place the caption at the top of the table. Place lengthy descriptions in the table footnotes. Tables should be laid out with clear row and column headings and units, where appropriate. Organize the table in such a way that readers can understand the contents clearly.

The tables should be self-contained and abbreviations/symbols used should be explained within the table in the footnotes. Avoid using unnecessary abbreviations and symbols. Include all details such as the unit(s), sample size, and statistical results in the table. Tables should be created within the document so that they can be edited by the editorial staff.

Figure Captions: Place a list of the figure captions after the tables. The figure captions should be informative. A figure should be self-contained and any abbreviations or symbols

used in the figure should be explained in the caption. When multiple figures are used in a figure, each figure must be explained clearly.
Figures: Submit figures as separate image files such as JPG and TIF. Figures must be of camera-ready quality. Pay attention to the relative proportion of the figure and the text used in the figure including the axis scales. Texts should be clear and legible. The design, size/ratio, and aesthetics of the figures matter.

HOW TO RESPOND TO REVIEWERS' COMMENTS:

The general principles for how to respond to reviewers' comments are:
1. Keep it professional, no emotional response! Authors should understand that, if taken properly, reviewers' comments (constructive criticism) can strengthen the manuscript substantially. After the initial reading, allow sufficient time to cool down. Start the revision when you are professionally ready.
2. Respond point by point to reviewer by reviewer (and the editor). Do not combine the responses. Highlight the revised section in the manuscript using colored text.
3. After clearly understanding the points raised by the reviewers, make your action plans point by point: to accept or to rebut. You do not have to accept/follow the reviewers' criticism/suggestion always. Start with minor items first and deal with the major items last.
4. Prepare the point-by-point responses in an elaborate manner. Avoid simple 'yes'- or 'no'-type answers. Reviewers put a significant amount of time and effort to read the manuscript to provide the comments and authors should honor their efforts by responding in an elaborated manner. Provide details of your revision within the response so that the reviewer does not have to read the manuscript again to find out exactly what changes were made. Copy and paste the relevant revised section into your response.
5. When you decided not to accept/follow the criticism/suggestions, clearly state the reason with supporting evidences. Even if you do not accept the criticism, try to revise the section of interest proactively to make your point clearer.
6. Asking the reviewers to understand your position or explaining your points to the reviewers alone is not sufficient. In most cases, it is the way the manuscript is written which raises an issue or causes a misunderstanding. Pay attention to where the question is coming from and revise the manuscript to clarify your points. Authors should ultimately answer to the reviewers' comments by revising the manuscript.
7. Do not omit any points raised by the reviewers. It will only guarantee one more round of revision at best.
8. If a point raised by a reviewer is not clear or not just in your perspective, check with the editor for clarification before you respond.

SUMMARY:

A set of general and section-specific writing principles for the manuscripts to be submitted to SB are presented in this paper. Complying with these principles will increase the manuscript's chance to survive the first round of the peer-review process. Understanding how to respond to the reviewers' comments is important as the manuscript's chance to be accepted for publication after the second round of the peer-review depends directly on how the authors respond to the reviewers' comments and revise the manuscript accordingly. The peer-review process is the ultimate testing ground for a research study and a manuscript is judged by how it is written and what it contains. A research study that fails to generate a publication is of no use as the study findings cannot be disseminated.

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ISBS 2010

Applied Session

Gymnastics

Applied Session
BIOMECHANICS AND GYMNASTICS

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The applied session at the 28th International Society of Biomechanics in Sport Conference focuses on two major themes of research; the **Coaching-Biomechanics Interface** and **Injury and biological loading**. These two interrelated themes underpin the understanding and knowledge needed to provide a safe and effective environment for the development of gymnastics skills and for the well being of performers. Ecological validity permeates these research approaches ensuring that meaningful information for coaches, scientists and clinicians is provided. The four presentations will use examples from evidenced based research on these themes. Two will focus on the coaching-biomechanics interface; one from an experimental perspective (Dr Gareth Irwin) and the other from a theoretical/modelling one (Dr Mike Hiley). The coaching-biomechanics interface is a term used to conceptualise how coaching can be informed from a biomechanical perspective. The process involved here is a continuous one, with each cycle starting and ending with the athlete. The process is based on a coach's tacit knowledge in relation to the practices that are routinely used to develop athletes' skills. Integral to this process, is the communication between the biomechanist and the coach and athlete. This cycle of extracting, processing and imparting new scientifically grounded knowledge and understanding represents the coaching-biomechanics interface. Sometimes this new knowledge may simply reinforce existing practices or it can provide new insights which inform future skill development. The overall purpose of developing the coaching-biomechanics interface is to bridge the gap between biomechanical science and sport practice. The interface aims to make training more effective and efficient for athletes who are already working near to their physiological limits.

The second theme presented by Dr Marianne Gittoes and Professor Peter Brüggemann examines the effects of gymnastic performance on the loading of the biological structures of the gymnast. These two talks will explore impact loading and the resultant physical demand on performers. Examples will include landing in gymnastics. One fundamental topic will be an examination of the ability of the gymnast to voluntarily i.e. consciously modify technique. Investigating what the gymnast can do and what is inherent in predisposing them to high loads therefore how does the gymnast interact technique changes with inherent mechanisms of load attenuation.

Presenters are:

1. Dr Gareth Irwin (Wales): Coaching Biomechanics Interface: Competition and training
2. Dr Mike Hiley (England): Coaching Biomechanics Interface: Simulation modelling
3. Dr Marianne Gittoes (Wales): Variability and performance: implications for injury in gymnastics
4. Prof Dr Peter Brüggemann (Germany): Biological load and injury in gymnastics

COACHING BIOMECHANICS INTERFACE: COMPETITION AND TRAINING

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KEYWORDS: gymnastics, bars, kinematics, kinetics, progressions, physical preparation

INTRODUCTION:

Coaching-biomechanics interface

Bridging the gap between the underlying biomechanical parameters that determine successful gymnastics performance, and the provision of meaningful information for coaches has been the challenge for sports biomechanists for decades. Conceptualising this fundamental relationship through the coaching-biomechanics interface draws on the cognitive processes of learning and understanding, combined with grounded scientific concepts, which help explain and increase understanding of gymnastic performance. As such the coaching-biomechanics interface begins with an examination of coaches' implicit knowledge highlighted through the conceptual models of skill learning and development (Irwin et al., 2005). Central to this model is the development of a mind set, a conceptual understanding of how a skill works. Coaches develop an understanding of how the skill works then aim to replicate the spatial and temporal characteristics of the final skill in the physical preparations, progressions and preparatory skills used in training.

Gymnasts are currently training close to their biophysical limits and with the evolving Code of Points (FIG, 2009) and desire to continually strive for more complex and innovative moments it is desirable to enhance training by using objective criterion against which skill development pathways can be measured. The ultimate aim of the coaching-biomechanics interface is to make training more effective, efficient and safe, incorporating the needs of the elite performer in parallel with considering the well being of the individual. The following two examples provide research-based evidence of the coaching-biomechanics interface employing biomechanical studies based on the fundamental principles of training to help understand the development of a key gymnastic movement (high bar longswing) and explain techniques of release and regrasp skills on uneven bars (Tkachev). Previous research in the area of high bar and gymnastics has been dominated by groups from Loughborough (Hiley et al., 2007; Yeadon and Hiley, 2000) and Cologne (Arampatzis and Brüggemann

Skill Development

A series of studies have been conducted which have resulted in novel biomechanically driven scores, based on the principle of specificity, and incorporating movement variability and difference which aimed to assess the effectiveness of progressions.

The biomechanical scores were applied in three ways; firstly examining the single joint orientation of the hips and shoulders (Irwin and Kerwin, 2005); secondly from a more holistic perspective examining the interaction of the hips and shoulders using measures of inter joint coordination, namely continuous relative phase (Irwin and Kerwin 2007a); and finally employing an inverse dynamics approach, scores were applied to the musculoskeletal demands placed on the performer (Irwin and Kerwin, 2007b). When considering kinematics and kinetics similarity it was suggested that progressions, which cause gymnasts to use similar levels of energy to the longswing, are placing a stress on the musculoskeletal system in a specific manner. Although the energy level, in a progression, may be similar, this does not always correspond to similarities in the movement pattern. As a consequence the physiological adaptations which occur through training may not be effective or desirable. Different classifications of progressions therefore exist with those that replicate the movement pattern (kinematics) and those that replicate the physical demand (work done/ energy expenditure). These studies have generated further questions for example, how is skill development effected by the choice of progression?

Understanding technique

Official changes to the rules governing the bar spacing on uneven bars in the late 1990's have enabled female gymnasts to perform different versions of the complex but common release and regrasp skill, the Tkachev. Historically the Tkachev has been performed with the gymnast facing outwards and travelling towards the low bar whilst clearing the high bar. Increased bar spacing has enabled females to longswing the opposite way, facing inwards and travelling away from the low bar when performing the Tkachev.

This change in direction highlights two issues; firstly relating to the scoring system (should both versions be valued with the same difficulty) and secondly, are these skills, which appear similar, placing the gymnast under the same physical and technical demands. The responses to these questions could have implications for the most effective physical preparations for a gymnast. As such a biomechanical investigation was carried out with the specific aim of quantifying the differences in musculoskeletal work between the outward and inward Tkachevs, and to examine whether these skills are equally demanding on gymnasts (Kerwin and Irwin, 2010). Based on the premise that gymnastics coaches visualise all skills as a series of shape changes and movement patterns (Irwin et al., 2005), the current study observed kinematic similarities in the two variants of the straddle Tkachev which masked differences highlighted by the subsequent kinetic analyses. In particular, the musculoskeletal work at the shoulders was found to be predominantly positive in the outward and negative in the inward variant of the skill. These differences underpin variations in rate of change in angular momentum with the inward variant being superior for generating improved release conditions and greater reversal of angular momentum. There are two implications of these findings. The first is that the inward version of the Tkachev provides the gymnast with the opportunity to produce more complex versions of the skill through alterations in body shape in flight (e.g. piked or straight body). Secondly, from a classic training principles perspective, in order to develop the inward Tkachev, gymnasts need to change the preparatory activities to elicit the specific musculoskeletal adaptations which correlate more closely with those required in the inward variant of the skill. This study has highlighted that apparent similarities in the kinematics mask fundamental differences in the kinetics and expands the ideas promoted by Irwin and Kerwin (2007b) when ranking progressions for skill development based on musculoskeletal demands.

SUMMARY

This paper has highlighted the coaching biomechanics interface as an integrated concept within the coaching process. The aim is to bridge the gap between the underlying biomechanical parameters that determine successful performance and the communication of this information in a meaningful way to coaches. We have considered how progressions can be organised based on biomechanical principles and also how techniques can be developed through enhanced understanding of key components of selected skills. This approach aims to make training more effective, efficient and safe at all levels of performance.

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NEUROMECHANICAL LOAD OF BIOLOGICAL TISSUE AND INJURY IN GYMNASTICS

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Biomechanical and biological factors related to the performance enhancement in gymnastics indicate that the potential of the neuro-musculoskeletal system seems to be close to or even at the ultimate tolerance limits. Acute and chronic severe tissue injuries are frequently reported from male and female gymnasts. In general mechanical loading of the musculoskeletal system is a prerequisite for morphological and functional adaptation of biological material. But if stress and strain increase to a certain level and exceed the mechanical limits of the individual structure, mechanical loading may lead to tissue damage. Young gymnasts have been shown to be particularly prone to overuse injuries, as their musculoskeletal system is still immature. One of the most serious overuse problems for young athletes is the development of abnormal radiological signs in the lumbar spine and the thoracic-lumbar transition. Research on biological load and injury in gymnastics indicated that tissue abnormalities, chronic injuries and acute tissue damage are related to mechanical load, the technical devices and apparatus; as well as, neuromuscular performance, muscular strength and technique.

KEY WORDS: mechanical loading, morphological adaptation, injury

INTRODUCTION: Mechanical load acting on the biological structures of the human musculo-skeletal system during artistic gymnastics is one possible stimulus to maintain and/or increase the strength of biological material. Excessive load may lead to microscopic or macroscopic damage of the anatomical structures. From this perspective mechanical overload is the cause of the damage of one or more biological structures, and an injury can generally be defined as the damage of biological tissue caused by physical loading. Injury can result from a single overload exceeding an individual tissue's maximum tolerance. Such a situation may lead to catastrophic spine injuries. The term catastrophic injury is defined as any injury incurred during participation in sport in which there is a permanent severe functional neurological disability (non fatal) or a transient but not permanent functional neurological (serious) disability. A chronic injury is initiated by microscopic damage of the tissue's structure. Long term repeated loading could worsen the injury eventually becoming macroscopic and/or resulting in tissue degeneration. Based on this definition a relationship between injuries and mechanical energy can be concluded. The principal relationship between mechanical energy and injury gives us reason to examine the causes of muscular-skeletal injury especially in sports where high amounts of mechanical energy and mechanical forces are a prerequisite for successful activity and performance. Gymnastics is a typical sport, which is intuitively combined with high loading of the spinal structures.
Some epidemiological studies suggest that gymnastics (Goldstein et al. 1991) may accelerate the degeneration of spinal structures. Sward et al. (1990) e.g. investigated 142 Swedish top athletes who competed in wrestling, gymnastics, soccer and tennis. All groups of athletes reported previous or present back pain at a higher frequency than found in previous studies of the general population. Radiological abnormalities of the thoracic-lumbar spine occurred in 36-55 % of the athletes. The results may suggest a causative relationship between athletic activities and radiological abnormalities. The radiological findings indicate both direct traumatic changes as well as disturbed vertebral growth. Sward et al. (1990) concluded that both the age at onset of athletic activity and the degree of mechanical load on the skeleton are important factors in the development of these abnormalities.
Only a few recent papers focus on gymnastic related tissue damage; this might be related to a decreased public interest in this sport even if the performance enhancement and the increase of technical difficulty in this dedicated sport exploded in the last two decades. One can speculate that the decreased public interest is related to reduced funding opportunities

for applied research in this area and that this causes the deficit of research and epidemiological papers.
Kujala et al. (1992) found some evidence that as intense physical activity in young athletes (10.3-13.3 years) increases the occurrence of low back pain (LBP). Several papers reported or speculated on a relation between mechanical loading in gymnastics and back injuries (e.g. Petrone and Ricciardelli 1987). The total time of exposure was identified to be the most important cause for increased disc degeneration (Goldstein et al 1991). Some authors warned about overloading the spine by physical activity and sports in the adolescence (Micheli 1985). Other studies found no higher frequency in disc degeneration in young gymnasts than in controls (Terrti et al 1990). Tsai et al. (1993) reported no differences in the frequency of self reported LBP between former gymnasts and controls.
Only few findings of mechanically induced injuries of discs and vertebrae in gymnastics and no data on the loading of the spinal structures during the different gymnastic skills and drills can be identified in the literature. From such a perspective the often-used cause-effect-relation between loading in gymnastics and spinal abnormalities is more intuitive and not well understood. On the one hand there are the positive effects of physical activity and sport in adolescents. On the other hand there are the noteworthy findings regarding the limited load capacity of the developing system. The dual role of mechanical loading as a positive and negative influence on the biological structures and of how beneficial effects of training interact with potential harmful effects of mechanical loading has received relatively little attention. Genetics may also be an important factor in the degeneration process. In a study on monozygotic twins Battié et al. (1995) demonstrated the importance of genetics in relation to a life time of physical activity with regard to disk degeneration. Therefore the individual genetically determined mechanical properties of the biological material should not be underestimated.
To approach a better understanding of the relationship between mechanical loading due to gymnastics and the damage or injuries of biological structures the frequency and severity of an injury or a group of injuries should be evaluated. From these figures it could be determined which factors lead to a particular damage or abnormality. Then the relation between these factors and the specific injury or abnormality should be examined to better understand the factors responsible for the tissue response.

METHODS: In a five years prospective study a total number of 135 female gymnasts (10-22 years) were surveyed. From these gymnasts fifty-seven elite female gymnasts were examined over at least three consecutive years. The highest frequency of spinal disorders was found in an age of 12-13years. During the survey 37 of the gymnasts were aged from 12.5 to 15.5 years and therefore in the most vulnerable phase of life. These 37 female athletes were chosen for further analysis. Thirty non-athletic females at the same age were used as controls. The clinical examination was performed twice during the year by the same well-trained medical staff using a precisely defined examination protocol. In cases with a clinical indication or a positive clinical finding radiographs were taken. Once a year, a MRI study of the gymnasts' lumbar spine and the thoracic-lumbar transition was performed.
Muscle strength and anthropometric variables of the subjects under study were measured several times each year. Mechanical loading was estimated from a biomechanical model using kinematic, kinetic and EMG measurements as model input during the most important movements and skills. Using the training protocols the cumulated spinal load for the day, the week and the total year was estimated. The resultant moments and forces at L5/S1 and/or Th12/L1 were calculated using inverse dynamic techniques. For the distribution of the resultant forces and moments to muscles, ligaments, and contact forces at the motion segment an EMG based switch technique was combined with optimization methods.

RESULTS AND DISCUSSION: Forty-nine positive findings in 135 gymnasts in the thoracic-lumbar transition support the weakness of the anterior part of the motion segment. Abnormalities in the posterior part of the ring apophysis are infrequent. From the positive findings in the prospectively surveyed sample (n=37) 59,4% of the signs disappeared or decreased in intensity during the three years control, in 40,6% of the cases the level of severity of the abnormalities increased. Hellström et al. (1990) reported a higher frequency of vertebrae with abnormal configuration (e.g. flattening, wedging and increased sagittal diameter) in young athletes than in non-athletes. They argued that healing of moderate vertebral fractures in children may be disturbed by high intensity loading and can explain the abnormal configuration. Our prospective data indicate some remarkable tissue responses to the induced loading. These responses are an increase of bone mass, an increase in plate area and an increase of normalized disk height and of water content in the intervertebral disks. Sward et al. (1992) found reduced disk MRI signal intensity more than twice as common in male (adult) gymnasts (mean age: 23 years) than in non-athletes. Hellström et al. (1990) reported disk height reduction in wrestlers and gymnasts in comparison with non-athletes. The cause for the different results to other studies (Tertii et al. 1990) may be that the young individuals in Tertti's sample had not passed the growth spurt, a period of time in which the growth plates and apophyses are most sensitive to trauma. The findings of our prospective study support Terttii's results and found no higher frequency of a reduced disc height in the young gymnasts in relation to the controls. Abnormal vertebrae configurations (flattening, wedging, increased sagittal diameter) are of higher frequency in athletes than in non-athletes (Hellström et al. 1990). Our data indicate a peak incidence at a lower level at about the thoracic-lumbar junction. The reported relation to back pain was not supported through our finding. A total of 47.4% of the 135 gymnasts never reported pain during the entire period of the survey. Sward et al. (1990a) concluded that their findings were highly suggestive of a causative relationship between rigorous athletic activities, radiological abnormalities, and back pain. The radiological findings indicate both direct traumatic changes as well as disturbed vertebral growth. Our clinical and radiological examination of former elite gymnasts supported Sward's data in principle and reported the majority of severe and moderate vertebral deformities in the thoracic-lumbar junction. About one third of the examined former female gymnasts (n=37) had severe vertebral deformities, another third showed moderate findings. The most frequent osteochondroses were found in the age groups of 12-15 years of age. These data of the examination of traumatic changes and abnormalities emphasize the vulnerability of the spine especially during the growth period. Both the age of onset of physical activity and the degree of mechanical load on the skeleton should intuitively be discussed as causal factors in the development of the described abnormalities.

The response to mechanical loading in the young gymnasts was shown in a significant ($p<0.05$) increase in vertebral bone mass in comparison to the controls at same age. An increase of BMD during the period of survey was identified in all the cases. In addition an increase of the vertebral endplate areas in the thoracic and the lumbar vertebrae was found. Surprisingly the water content of the disks of the young gymnasts (9 to 13 years) was significantly higher than in the controls. This finding corresponds with a moderately greater disk height of the gymnasts in comparison to the controls. In the older gymnasts (14 to 19 years) a first tissue ageing was identified. It is of interest that age of the material (even in the young gymnasts) of the disk explained 22.4% of the variance of the disk Tw2 signal (water content) whereas only 11.2% was explained through the mechanical loading. 66.4% of the Tw2 variance could not be explained by load or age. This indicates that genetics should not be underestimated when considering biomaterials strength and tissue response to load.

CONCLUSION: The mechanical load of the spine in gymnastics may reach or be close to the limits of tissue tolerances. The injury or the tissue damage will occur when these limits are exceeded in one traumatic failure or in repeated micro-failures. Apparatus, gymnastic technique and training determine the mechanical load of the biological structures. In order to prevent spinal injuries these factors have to be discussed to reduce the risk that loads exceed the physiological limits. The proper technique can contribute to the reduction and control of mechanical load of the spine. This is especially valid for the posterior column of the lumbar spine, the lumbo-sacral junction, and the anterior column of the thoracic-lumbar spine. Forward bending or inclining the trunk when high external forces are applied to the

body lead to an increase of peak pressure on the anterior area of the vertebrae at the thoracic-lumbar junction. This mechanism loads the apophysical ring when landing from a jump or a dismount with forward lean and bent trunk position. The landing technique and the technique of the skills prior to landing have a major influence on the pelvis and spine posture and therefore on peak pressure and pressure distribution at the motion segment. In bending with lumbar spine flexion up to 45° to horizontal the trunk is counterbalanced by the erector spinae muscles. If the flexion increases to about 60° the passive posterior ligaments become taut. In continuing the forward movement the pelvis rotates forward until the pelvis rotation is passively restricted by the gluteus and hamstring muscles. In this deep forward lean position no muscular activity of the back muscle can be registered. The inertial force and the weight of the trunk is counterbalanced by passive forces of ligaments, fasciae and muscles. Therefore landings with an extreme forward lean contribute to the load of the passive posterior structures. The landings with a more or less upright trunk position are controlled by the erector spinae muscles, which must clearly be activated prior to landing. In order to minimize the total muscle force and the resulting compression force of the motion segment well prepared athletes do not co-activate the abdominal muscle in landing.
The physical preparation of the athlete is a major concern in the prevention of spine injuries. This is true for expected loading and falls. The development of a muscular corset for the thoraco-lumbar spine seems to be the most effective strategy to ensure a controlled mechanical loading. The strengthened back muscles are able to control and counterbalance the trunk's weight and inertial force during impact loading.

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COACHING BIOMECHANICS INTERFACE: SIMULATION MODELLING

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KEY WORDS: optimisation, virtual reality, co-ordination.

INTRODUCTION: Computer simulation modelling is a powerful tool that can be used to gain insight into the mechanics of gymnastics technique. Having identified the underlying mechanics performance may be optimised and new skills developed. As computer processing power has increased so have the applications for simulation modelling. Research is currently being conducted to combine simulation models with virtual reality environments to produce innovative gymnastics training aids.

PARALLEL BARS: In gymnastics there is often more than one technique used to perform the same skill. In these situations the question is "Which is the best technique?". The 'best' technique will depend on many factors including the age and preparation of the gymnast and the long term plan of the coach (i.e. which technique will allow the development of more complex skill variations). Simulation modelling and optimisation can be used to help answer such questions. Two techniques were identified in the coaching literature for the undersomersault to handstand on parallel bars (Davis, 2005). Two optimisations were performed with the score based on (a) minimising joint torques and (b) requiring a vertical path of the mass centre prior to release (Hiley, Wangler & Predescu, 2009). The two optimal solutions closely resembled the techniques identified in the coaching literature (Figure 1). Both techniques may be used by the coach, but at different stages of the gymnast's development. The back clear circle technique (Figure 1a) is less demanding in terms of strength, and may be adopted during the initial stages of learning. Whereas the stoop stalder technique (Figure 2b), which requires more strength, has advantages for skill progression, for example in the undersomersault to handstand with either a half or full turn.

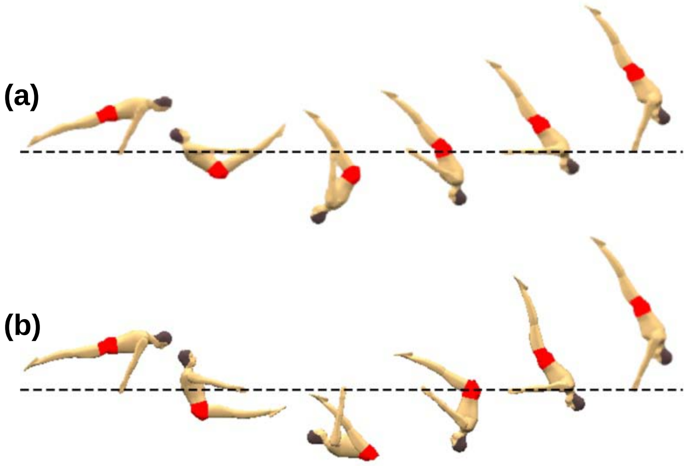

Figure 1. The optimal (a) back clear circle and (b) stoop stalder undersomersault to handstand techniques.

TWISTING SOMERSAULTS: Simulation modelling can also be used to improve performance, whether it is to increase the number of somersaults/twists that can be performed in a dismount or to predict totally new skills. Using a simulation model of aerial movement Yeadon (2009) addressed the following question posed by a national freestyle aerial skiing coach "How can five twists be produced in a triple somersault using asymmetrical arm movements?". The coach was very specific regarding the constraints placed on the technique, so that minimal changes to the take off and landing phases would be required. Alternative solutions were possible, but would have contradicted the take off technique preferred by the coach. In addition to developing a technique that produced five twists (Figure 2), a set of lead up skills was developed that would allow the athlete to safely build up to the final complex skill. The advantage of this approach is that progressions of skills can be developed with the model, in safety, prior to working with an athlete.

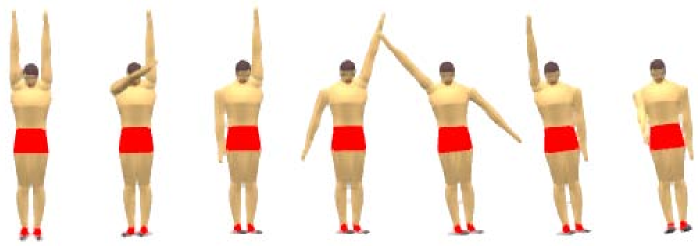

Figure 2. Arm actions required to produce five twists in a triple somersault.

HIGH BAR: When optimising performance gymnast-specific strength limits are normally used to prevent solutions exceeding the strength characteristics of the gymnast (King & Yeadon, 2004). However, optimum technique is often sensitive to timing errors, producing sub-optimal performances when perturbed (Hiley & Yeadon, 2008). Twenty giant circles prior to release for a Tkatchev were analysed to determine the level of timing variability at the instants of maximum and minimum hip and shoulder flexion and extension. The mean and standard deviation were calculated for each measure. The range of deviations from the mean ranged between ± 30 ms with a standard deviation of approximately 12 ms. Optimisations including the sensitivity of the technique to variations in timing have been shown to produce optimum technique closer to the actual performance of the gymnast (Hiley & Yeadon, 2007). Hiley & Yeadon (2005) used a simulation model of the gymnast and high bar to investigate the feasibility of performing a triple layout somersault dismount from the high bar. Starting from the 2000 Olympic high bar champion’s giant circle technique and using appropriate strength limits it was found that the model could produce sufficient angular momentum and time of flight to produce a triple layout somersault dismount. However, the time window within which the model could release the bar and successfully perform the dismount was unrealistically small compared to double layout somersault dismounts (Hiley & Yeadon, 2003). When the release window and timing variations were incorporated into the optimisation (Hiley & Yeadon, 2008) the model was only able to produce sufficient angular momentum to perform a triple piked somersault dismount (Figure 3). Since gymnasts are not able to precisely co-ordinate skills each time, the techniques they use will have been developed to cope with the level of timing variation present in the system. If optimisation is to produce realistic solutions that can be reproduced in the gymnasium by the gymnast considerations regarding the limits of co-ordination must be included.

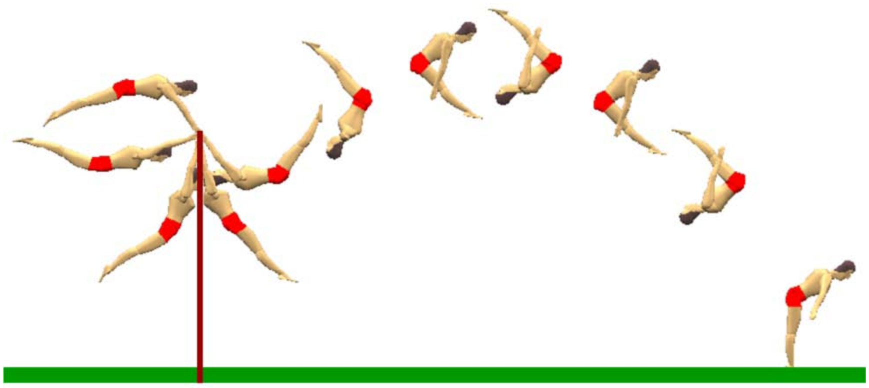

Figure 3. The triple piked backward somersault from high bar.

VIRTUAL REALITY: Simulation models have been combined with a head mounted real-time virtual reality display in order to train viewing skills during aerial movement (Yeadon & Knight, 2006). It might be expected that training gymnasts to view the landing area during flight may lead to more consistent landings. The system incorporates a three dimensional motion sensor attached to the gymnast's head which provides real time input to the simulation model. The view of the virtual environment from the model's head is then fed back to the gymnast through the virtual reality display. By programming the simulation model with the appropriate skill the gymnast can learn to view the landing area during the aerial phase before attempting it in the gymnasium. The system was tested on an experienced artistic gymnast who was about to learn a new skill on high bar (a double layout somersault with a full twist in the second somersault). The gymnast had not previously attempted to view the landing during high bar dismounts. After learning the appropriate head movements the gymnast undertook a progression of twisting somersault skills in the gymnasium, resulting in the desired high bar dismount (Figure 4). The gymnast provided positive feedback that the virtual reality training system had helped him learn to view the landing area during the new skill.

Figure 4. Gymnast using head movements to view through a twisting double somersault dismount.

CONCLUSION: The main benefit simulation modelling can provide gymnastics coaches with is an understanding of the underlying mechanics of specific movements. It is important that the knowledge gained from gymnastics research finds its way into coach education resources and practical application. It is also important that gymnastics research has input from coaches so that the right questions can be asked and addressed.

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THE PHYSICAL DEMANDS OF GYMNASTIC-STYLE LANDINGS: UNDERSTANDING AND ALLEVIATING INHERENT PREDISPOSITION

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The research aims to develop insight into inherent mechanisms and regulatory strategies contributing to the physical demands of gymnastic-style landings. The use of a modelling approach to examine the interaction of: 1. a performer's physical profile and self-selected landing strategy and 2: local mass distribution and mass tuning effects on impact loading is presented. Strategy adjustments accommodating inherent physical profiles were found to be essential in ensuring effective load attenuation but were acknowledged as potentially incompatible with current constraints in gymnastic scoring systems. Mass tuning partially alleviated the loading effects of inherent local mass profiles and was considered achievable without substantial alterations in the regulatory movement patterns.

KEY WORDS: physical profile, mass tuning, landing strategy, simulation

INTRODUCTION: Biomechanical research has supported the notion that the large and rapid forces incurred in landings frequently performed in gymnastic routines impose a predisposition to musculoskeletal injury (Daly et al., 2001). As highlighted by Decker et al. (2003), it is generally accepted that the internal and external loads experienced in landing may be manipulated by the lower extremity kinematics or movement patterns used. Traditional laboratory-based studies have suggested regulatory control strategies such as the use of greater (Salci et al., 2004) and more rapid knee and hip flexion (Yu et al., 2006) to assist the execution of safe and efficient landing manoeuvres. In addition to control strategies, multiple inherent mechanisms have been considered influential in the impact loads experienced and the potential for lower extremity injury. Anatomical, neuromuscular and hormonal factors have been reviewed as traditional contributors to lower limb injury in sports performers (Boden et al., 2009). More recently, soft tissue mass properties (Liu & Nigg, 2000, Pain & Challis, 2006) and lower body stiffness (Butler et al., 2003) have been recognised as important contributors to the forces incurred during sport-related impacts.

Given the possibility of an inherent predisposition to excessive mechanical loading, Daly et al. (2001) highlighted recent discussions for rule changes to de-emphasise 'sticking' landing routines in the scoring of gymnastic dismounts. Allowing self-selected landing strategies may accommodate a conscious or instinctive compensation for inherent predispositions to the high physical demands incurred in landing. Gaining an insight into the contribution of inherent mechanisms to mechanical loading and the potential influence of regulatory strategies may be valuable for developing understanding of injury predisposition and in supporting customised landing responses in the scoring of gymnastic landings.

Due to control constraints associated with a traditional laboratory setting, the interactive and independent effects of inherent load attenuation and regulatory mechanisms such as landing strategy changes has been difficult to ascertain. Biomechanical modelling offers a contemporary alternative to traditional experimental investigation due to the ability to isolate and readily manipulate the mechanisms under investigation. The aim of this research was to subsequently gain an insight into inherent mechanisms and potentially modifiable strategies influencing the physical demands experienced in gymnastic-style landings using a contemporary biomechanical modelling approach.

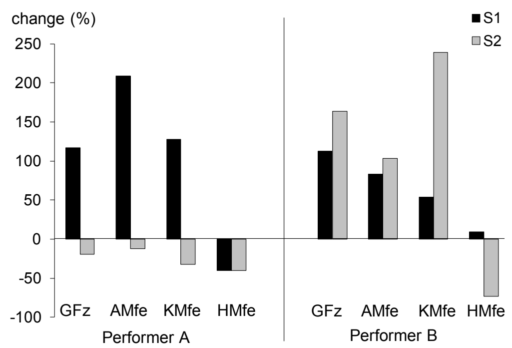
METHODS: A four-segment, non-rigid simulation model incorporating soft (wobbling) and rigid masses was developed and evaluated using the procedures presented by Gittoes et al. (2006). The evaluated simulation model was used to replicate customised, gymnastic-style drop landings (height 0.46m) for two female performers (A: body mass 56.8 kg; B: body mass 69.0 kg). The self-selected strategy used in the simulated landings was defined by performer-specific ankle, knee and hip joint kinematic profiles derived from actual landing performances. The visco-elastic properties of the foot-ground interface were represented using spring-damper systems that replicated the ground reaction forces incurred in the simulated landings. Customised visco-elastic properties representing the coupling between wobbling and rigid masses were used to represent mass tuning responses produced in the motions. The simulation model was subsequently employed to investigate mechanisms contributing to the physical demands incurred in the gymnastic-style drop landings.

Application 1- Influence of inherent physical profiles and landing strategy: Personalised whole body physical profiles derived using a component inertia model (Gittoes & Kerwin, 2006) were integrated with the non-rigid simulation model to produce motions executed using an independent performer's landing strategies. Secondly the personalised physical profiles were used to replicate landings executed with the performer's own self-selected and unselected landing strategies. Comparisons of the impact loads produced in the customised evaluated motion and the motions simulated using modified (independent performer and own unselected) landing strategies were subsequently made.

Application 2- Influence of inherent physical profiles and mass tuning: Personalised segmental wobbling and rigid mass distributions (Rm_p) and mass coupling (stiffness: k_{WR} and damping: c_{WR}) properties used in the customised evaluated motion were simultaneously modified by ±5% perturbations. Perturbations in the Rm_p were made in combination with k_{WR} and c_{WR} changes. A positive perturbation in the Rm_p, coupling k_{WR} and c_{WR} produced a larger segmental rigid mass distribution and increased coupling stiffness and damping, respectively. The maximum vertical ground reaction force (GFz) effects caused by simultaneous Rm_p and k_{WR} and c_{WR} modifications were compared to the maximum GFz changes reported in Gittoes & Kerwin (2009) as a consequence of independent Rm_p modifications (±5% perturbations).

RESULTS:

Effects of inherent physical profiles and landing strategy: The use of an independent performer's strategy incurred an attenuated maximum GFz, and ankle and knee joint flexion-extension moments than the self-selected strategy for one performer (B) and had inconsistent loading effects on the remaining performer (A) (Figure 1a). The use of an independent performer's strategies had the greatest effects on the maximum knee joint flexion-extension moment (113 ±93% mean difference from self-selected response) when compared to the GFz (103 ±60%) and ankle (102 ±81%) and hip (41 ±26%) joint flexion-extension moment.



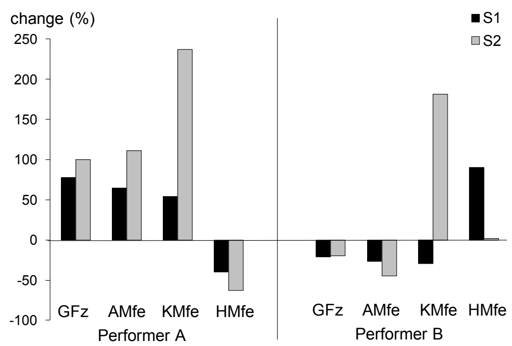
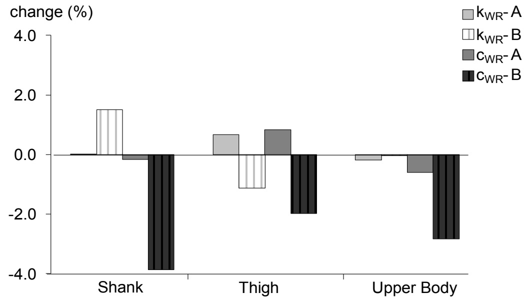


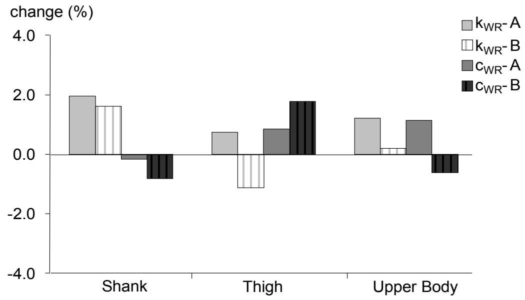
Figure 1: Change (%) in the maximum GFz and ankle (AMfe), knee (KMfe) and hip (HMfe) joint flexion-extension moment incurred using an (a) independent performer's and (b) the performer's own unselected landing strategies. [S1: strategy 1; S2: strategy 2]. Adapted from Gittoes & Kerwin (2008).

The use of a performer's own, unselected landing strategy incurred larger maximum GFz and ankle and knee joint flexion-extension moments for one performer (A) (Figure 1b) and typically attenuated effects for the corresponding measures of Performer B. A modulated personal strategy had the greatest effect on the maximum knee joint flexion-extension moment experienced (126 ±100% mean difference) but had relatively less effect on the impact loads (73 ±31% mean difference across all measures) compared to the use of another performer's strategy (90 ±29% mean difference).

Effects of inherent physical profiles and mass tuning: With reduced Rm_p, simultaneous c_{WR} increases typically produced larger changes in the maximum GFz experienced in the drop landings compared to simultaneous k_{WR} reductions (Figure 2). The notably attenuated maximum GFz incurred with independent upper body Rm_p reductions were further attenuated by increased c_{WR} by as much as 0.03 BW and 0.13 BW for landings performed by Perfomer A and B, respectively (Figure 2a).
The interaction of reduced Rm_p and k_{WR} had a relatively smaller but consistent effect (Figure 2b) on the maximum GFz compared to reduced Rm_p and increased c_{WR}, which produced idiosyncratic effects on the maximum GFz changes incurred with independent Rm_p modifications.



a.



b.

Figure 2: Interaction effects of Rm_p and k_{WR} and Rm_p and c_{WR} on GFz_{max}. (a) -5% perturbations in Rm_p and k_{WR}, and +5% in c_{WR}; (b) +5% perturbations in Rm_p and c_{WR}, and -5% in k_{WR}. Maximum GFz changes are reported relative to the maximum GFz changes incurred with corresponding independent Rm_p modifications. Adapted from Gittoes & Kerwin (2009).

DISCUSSION: An inherent predisposition to large and rapid mechanical loads in gymnastic-style landings potentially exists. The interactive contributions of a performer's inherent physical profile and regulatory control mechanisms to impact loading was investigated and may support alleviated constraints in the scoring of gymnastic dismounts. The simulation modelling approach used successfully allowed an examination of the isolated effects of the inherent and modifiable mechanisms under investigation.
The heightened impact loads experienced when using another performer's strategy or an unselected personal strategy suggested instinctive or conscious customisation of landing strategies to an inherent, whole-body physical profile and movement conditions. The greater impact load sensitivity in the landings simulated with another performer's strategy compared to the unselected personal strategy, further suggested strategies customised to inherent physical profiles may be prioritised over adjustments to diverse landing manoeuvres for effective impact load attenuation.
The whole body mechanism of landing strategy selection was found to be capable of influencing the maximum GFz incurred by a predisposing physical profile by as much as 103%. However, localised mass tuning adjustments were also found to provide a notable but smaller contribution to impact loading (up to a 3.9% change in maximum GFz). As highlighted in Gittoes and Kerwin (2009), reductions in the damping between soft (wobbling) and rigid masses were found to positively interact with lower, local rigid mass proportions, particularly in the upper body, to help further attenuate the impact loads experienced during the simulated landings. Liu and Nigg (2000) previously supported the notion that through muscle tuning, mass coupling properties may interact with inherent mass distributions in the body to control the impact forces incurred in less dynamic running ground contact phases. Without modifications to the kinematic landing strategy employed, mass tuning achieved by

developing and modulating neuromuscular responses may provide an alternative mechanism for alleviating the high impact loads naturally incurred by a performer's physical profile.

CONCLUSION: Considering the likely maintenance of a scoring system for gymnastic dismounting that requires constrained movement patterns, achievable modifications in mass tuning may alleviate the physical demands experienced in gymnastic-style landings without substantial alterations in the movement patterns produced. However, accommodating self-selected landing strategies that adjust to diverse physical profiles and movement conditions in the scoring system may provide substantially greater protection benefits for performers executing the potentially injurious manoeuvres.

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COMPARISON OF SPRINTING MECHANICS OF THE DOUBLE TRANSTIBIAL AMPUTEE OSCAR PISTORIUS WITH ABLE BODIED ATHLETES

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The purpose of this study was to examine the overall kinetics and the kinetics at the joints of the lower limb while sprinting at maximum speed, and to compare the data of a double transtibial amputee (OP) and able-bodied controls running at the same level of performance. One double transtibial amputee, and five able-bodied sprinters participated in the study. The athletes performed submaximal and maximal sprints on an indoor track embedded with 4 Kistler force plates while recorded with a 12 camera Vicon 624 system. OP displayed lower mechanical work (stance phase), external joint moments and joint power at the hip and the knee joints while displaying higher values of joint power at the (prosthetic) ankle joint compared to able-bodied athletes. The mechanical work at the knee joints was 11 times higher in the negative phase and 8.1 times higher in the positive phase during stance in the able-bodied athletes compared to OP.

INTRODUCTION: On behalf of the International Association of Athletics Federations (IAAF) the biomechanics of maximal sprinting of the double below knee amputee Oscar Pistorius was analyzed in the phase of individual maximal running speed. In comparison to that, the sprinting mechanics of five able bodied athletes of a similar level of performance were analyzed. IAAF used that information for the decision making process whether or not Oscar Pistorius was allowed for participating at the Bejing Olympics 2008.

METHOD: Five able bodied sprinters were recruited as control group (CTR). Their average 400 m personal bests varied from 46.5 to 49.26 s, their average body mass was 78.67 ± 7.9 kg and, height 1.88 ± 0.05 m and their average age was 25.7 yrs. The personal best of the 22 years old double transtibial amputee (OP) was 46.3 s, his body mass was 83.3 kg and his body height was 1.85 m at the time of the measurement both with his dedicated sprinting prostheses (Cheetah, Össur, Iceland). Subjects performed maximal sprints over an indoor track of about 70 m. Ground reaction forces were measured using four 90 cm x 60 cm force plates (Kistler, Switzerland). Movement analysis was performed during stance phase. Sprinting kinematics were recorded using a twelve camera infrared high speed system (Vicon, UK). Segment kinematics of the lower extremities and inverse dynamic calculations were done using a three segment rigid body model (Stafilidis and Arampatzis, 2007). Therefore markers were placed on different anatomical landmarks of the able bodied sprinters and related points of the prosthesis (figure 1). Particularly joint power of the ankle and knee as well as the power generated and absorbed in the prostheses were calculated my multiplying angular velocity with joint torque component-by-component. Integrating the power time histories provided mechanical work produced in the particular structure.

Figure 1. Representation of the experimental setup. Left: Indoor running track, camera system and force plates. Middle and right: Marker placement.

A more comprehensive description of methods and material used in the measurement can be found elsewhere (Brüggemann et al., 2008).

RESULTS: Kinetics and kinematics of the sprints show clear differences between the able bodied group and OP. Figure 2 shows the vertical and anterior-posterior components of the ground reaction forces. Higher peak vertical and horizontal forces were found in the able bodied runners compared to the amputee. In addition the vertical as well as the horizontal breaking and propulsive impulses of OP were about 15% smaller than the impulses of CTR.

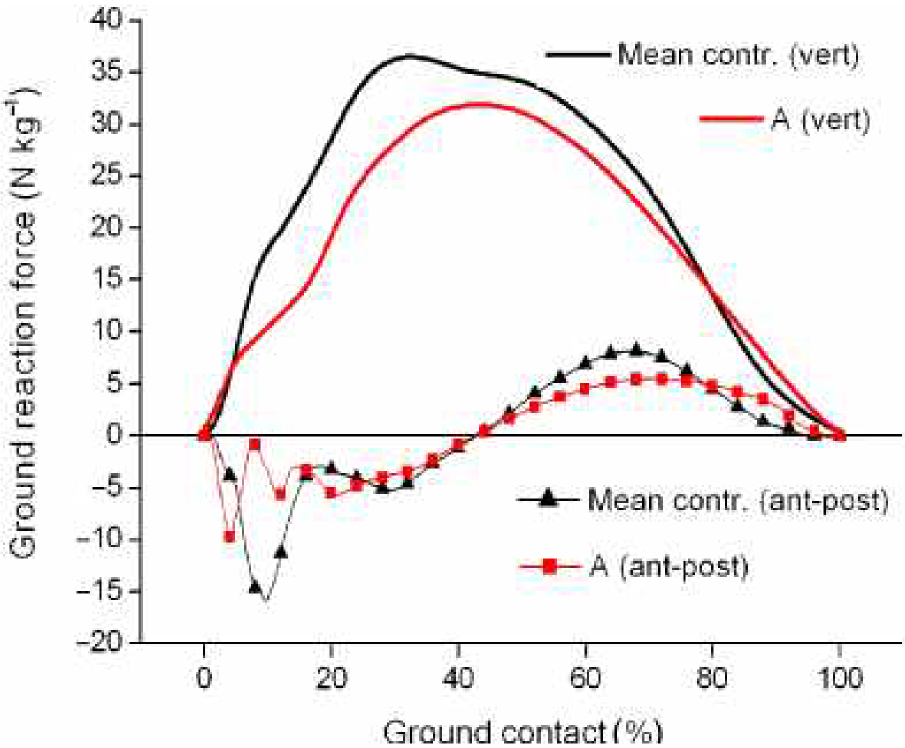

Figure 2. Vertical and anterior-posterior ground reaction force components for able bodied athletes (black) and OP (red).

Joint power and joint work in the prosthesis was much bigger compared to the ankle joint of the able bodied population (figure 3).

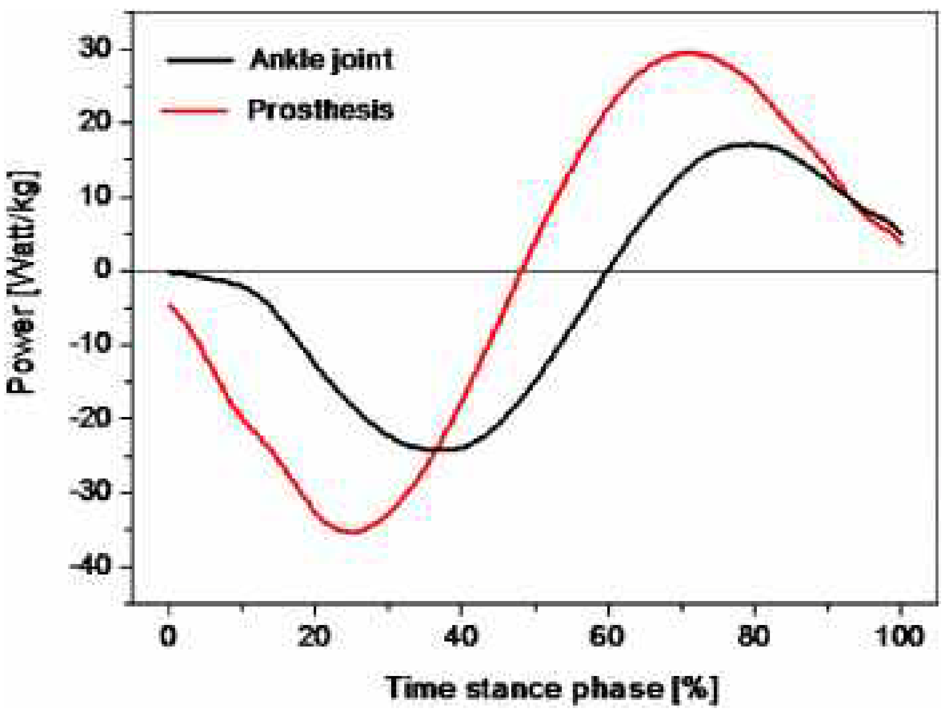

Figure 3. Time history of the joint power of the prosthesis of OP (red) and the ankle (black) of the able bodied controls.

The energy absorption of OP is about 40% higher for the artificial ankle joint compared to CTR. The almost elastic prosthesis keel returns about 90 to 95% of the stored energy. In the able bodied ankle joint only about 40 to 45% of the absorbed energy is generated in the second phase of the stance. The opposite situation occurs at the knee joint. Here OP does hardly any work, while the knee contributes remarkably (about 50% of the ankle) in CTR.

DISCUSSION: The biomechanics of double amputee sprinting shows differences to able bodied sprinting. The ground reaction forces indicate that OP runs with a smaller vertical displacement (smaller vertical impulse) in the phase of maximal speed than his able bodied counterparts. In addition he decelerates less in the first part of the stance phase and therefore has to generate a smaller propulsion impulse in the second phase of stance. The major part of the work of the lower extremity is done in the ankle joint in OP while the knee joint is contributing with less than 5%. This is completely different in able bodied sprinting, where the knee joint has a considerable contribution mechanical work production.

REMARK: Based on this and some additional data, IAAF decided to exclude OP from the Olympic Games 2008 in Beijing. The Court of Arbitration of Sports (CAS) overruled that decision essentially due to juristic reasons. OP did not participate in Beijing because he did not fulfil the qualification criteria for 400 m sprint.

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CORRELATION BETWEEN FUNCTIONAL CLASSIFICATION AND KINEMATICAL VARIABLES IN ELITE WHEELCHAIR RUGBY PLAYERS

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KEYWORDS: kinematics, wheelchair sport, quad rugby.

INTRODUCTION: Wheelchair rugby is a Paralympic team sport for athletes with disabilities affecting the four limbs. Players are classified according to their functional level from 0.5 (lowest function) to 3.5 (highest function). A player's classification is based on muscle tests designed to evaluate the strength and range of motion of the upper limbs and trunk and also includes observation of the athlete on court (IWRF, 2008). Although the sport class is based on movement potential associated with neuromuscular function and performance of tasks related to the sport, it is not well known how functional classification in rugby correlates with variables strongly related to performance such as distance covered. In a previous investigation (Sarro et al., 2010), kinematical variables were analyzed in an international rugby competition and suggested a relation between functional classification and distance covered during the game. To further examine this relationship, this project aimed to investigate the correlation between functional classification and player physical performance as measured by distance covered during a game. In addition, the correlation was examined for each game quarter and as a function of velocity range.

METHOD: Video images of the 2008 Demolition Derby international wheelchair rugby tournament (Birmingham, AL) were obtained by two Basler cameras (4.0-12mm 1:1.2 1/2 CCTV) fixed at approximately 7.9 m above the court. The video images were captured, measured and visualized using a purpose-built interface. The positions of players who played an entire game (n = 18) during the tournament were determined simultaneously at 10 Hz. The players (36.9$\pm$5.7 years old) had spinal cord injury or were quad amputee. One player had classification 0.5, four 0.1, two 1.5, four 2.0, two 2.5, four 3.0 and one 3.5. The data were obtained with a tracking method based on computer vision techniques recently applied to characterize soccer player performance (Barros, et al., 2007). The distances between 44 points on the court were used to calibrate the cameras. The calibration parameters and the position of the players in the video sequences were used to reconstruct the 2D coordinates of each player using the Direct Linear Transformation method. Before analysis the 2D coordinates of the players' trajectories were filtered with a Butterworth low-pass zero-phase digital filter with a cut-off frequency of 0.4 Hz, determined by spectral analysis. For each player the following were computed: a) total distance covered during the game; b) distance covered during each quarter; c) distance covered in four different ranges of velocity (V1: $0 \leq V1 < 1.37$ m/s, V2: $1.37 \leq V2 < 2.74$ m/s, V3: $2.74 \leq V3 < 4.11$ m/s, V4: $4.11 \leq V4 < 5.5$ m/s). The distance covered was calculated as the cumulative sum of player displacement between two successive samplings. The association between the variables described above and the classification level of the players was determined by Pearson's correlation coefficients. A hypothesis test at $p < 0.05$ was applied to verify significance of each correlation.

RESULTS: Table 1 shows the correlation results and corresponding *p* values. A moderate significant correlation (*r* = 0.6) was found between functional player classification and total distance covered. Regarding player classification and distance covered in the four ranges of velocity, a significant correlation was found for the velocity ranges V3 (2.74 to 4.11 m/s) and V4 (4.11 to 5.5 m/s) during the total game and quarters 1, 2 and 4. No significant correlations were found in the velocity ranges V1 and V2.

Table 1. Pearson correlation coefficient *r* (*p*) between functional player classification and total distance covered (TD) during each quarter, during total game, and in each range of velocity [V1: 0≤V1<1.37 m/s, V2: 1.37≤V2<2.74 m/s, V3: 2.74≤V3<4.11 m/s, V4: 4.11≤V4<5.5 m/s].

	Quarter 1	Quarter 2	Quarter 3	Quarter 4	Total Game
TD	0.57 (0.01)*	0.58 (0.01)*	0.58 (0.01)*	0.63 (0.00)*	0.62 (0.01)*
TD in V1	-0.33 (0.18)	-0.27 (0.28)	-0.19 (0.45)	-0.22 (0.38)	-0.31 (0.21)
TD in V2	0.05 (0.85)	0.05 (0.85)	-0.06 (0.82)	0.12 (0.62)	0.05 (0.85)
TD in V3	0.65 (0.00)*	0.79 (0.00)*	0.21 (0.40)	0.75 (0.00)*	0.70 (0.00)*
TD in V4	0.49 (0.04)*	0.58 (0.01)*	-0.09 (0.72)	0.56 (0.02)*	0.48 (0.04)*

* significant correlation ($p < 0.05$)

DISCUSSION: In general, moderate to strong correlations were found between the classification level of wheelchair rugby players and distance covered during a game, however this relationship was not detected at lower velocities. The distance covered during more demanding situations (i.e., when traveling at a velocity over 2.7 m/s) was strongly correlated to player functional classification, suggesting that the distance covered can be related to the functional ability for wheelchair propulsion. These results provide evidence for the use of distance covered during a game at different velocities as an auxiliary tool in the classification process of wheelchair rugby players. Support for the use of kinematical variables as part of player classification has previously been shown for other Paralympic sports, like shot-putting (height and angular speed of release) and basketball (shooting mechanics) (Chow et al., 2000; Malone et al., 2002). Adding quantitative evaluations to the traditional method of classification would lead to a more evidence-based system.

CONCLUSION: There is a strong correlation between the functional classification of wheelchair rugby players and the distances covered during a game especially at higher velocities, providing evidence to support the use of kinematical variables in the sport specific classification system.

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A KINEMATIC ANALYSIS OF TRUNK ABILITY IN WHEELCHAIR FENCING: A PILOT STUDY

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The purpose of this study was to explore the trunk ability differences between category A and B participants in wheelchair fencing. The result showed that category B participants might perform similar performance as category A participants in maximum lunge velocity, maximum lunge angle and maximum fast return velocity. This result may provide information to International Wheelchair Fencing Committee (IWFC) for the need of research on Wheelchair Fencing Classification (WFC) to clarify the differences between these two categories of participants.

KEY WORDS: Wheelchair fencing, Disabled sports, Classification, Paralympics.

INTRODUCTION: Classification in disabled sports is referred to the way in which athletes are grouped for competition and is different from classifications in able-bodied sports (Tweedy, 2002; Vanlandewijck, 2006; Vanlandewijck & Chappel, 1996). For example in able-bodied sports, power lifting is classified by body weight, master runners are classified by age, but classification in disabled sports is based on an athlete's ability and disability characteristics. The objective of classification is to provide valid tests or assessments for grouping athletes with a disability or multiple disabilities and make each sport on a 'level playing field'. Classifications for disabled sports aim at promoting equitableness and fair competition. However, in the recent year, there was controversy about the validity of classification system among the Paralympics sports (Firth, 1999; Gil-Agudo, Del Ama-Espinosa, & Crespo-Ruiz, 2010; Tweedy, 2003; van Eijsden-Besseling, 1985; Vanlandewijck, et al., 2004).
Wheelchair fencing has beenan official sport of the Paralympics since the first Olympic Games for the disabled athletes in Roma 1960, and includes two functional classes among three weapons during each summer games. As many other Paralympics sport, WFC is a point scored system and provides a guideline to the classifiers to evaluate the ability of the wheelchair fencers, in which a fencer has higher score, he/she will be classified into a higher category (IWFC, 2009). Therefore, fencers who are in the higher category should have better trunk abilities than the lower category fencers. In the WFC system, classifiers required to assess the wheelchair fencer's trunk strength, range of movement and balance according to six functional tests which imply that is a key factor to identify fencer's ability (IWFC, 2009). Hence, the present investigation attempted to apply biomechanical methods to determine the difference of the trunk movement abilities (trunk angle and trunk speed) between the two category groups (category A and B)

METHOD: Eight male and six female Hong Kong elite wheelchair fencers participated in this study (Table 1). They all had over 3 years of fencing experience and five males and four females belonged to the category A, while three males and two females were in the category B. These classifications were based on their participation on previous international competition which was approved by the IWFC.

During the test, the subject was required to perform a lunge toward (attack) the tester and a fast return (defence) away from the tester with maximum speed (Figures 1 and 2).

Table 1. Demographic information

Fencer ID	Gender	Category	Paralympics experience	Diagnosis
1	M	A	No	AP
2	M	A	No	AP
3	M	A	Yes	PARA
4	M	A	Yes	PO
5	M	A	No	PARA – W
6	M	B	Yes	PARA – W
7	M	B	Yes	PARA – W
8	M	B	Yes	PARA – W
9	F	A	Yes	AP
10	F	A	Yes	HEMI
11	F	A	Yes	PARA – W
12	F	A	Yes	HEMI
13	F	B	Yes	PARA – W
14	F	B	No	PARA - W

AP: Amputee, PARA: Paraplegics, PO: Polio, W: Wheelchair bound, HEMI: Hemiplegic


Figure 1: Lunge (attack)

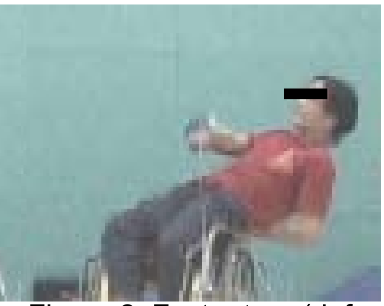
Figure 2: Fast return (defence)

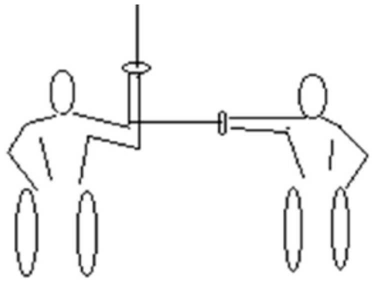
Figure 3: Fencing distance normalization

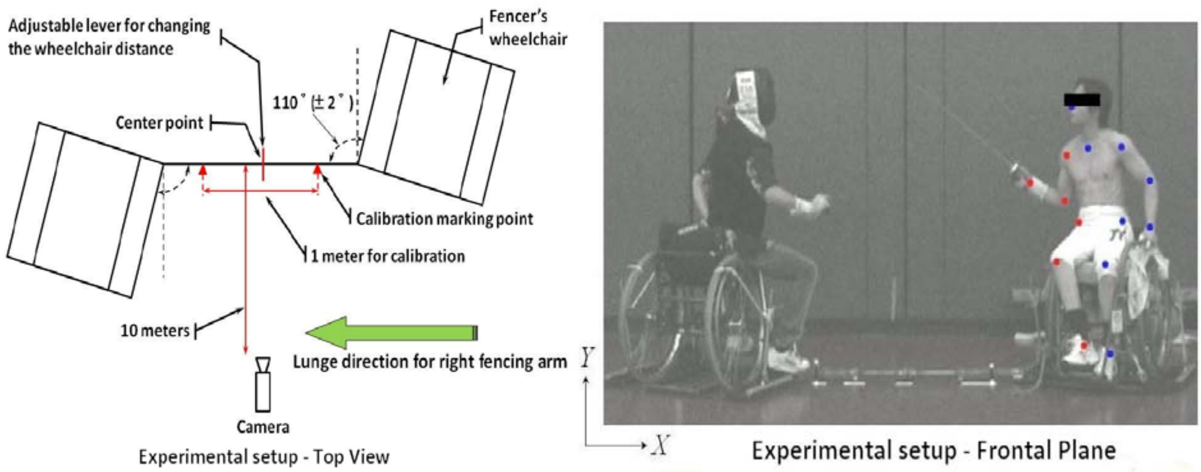

Figure 4: Experimental setup

There were five trials per movement with only the fastest being used for statistical analysis. Moreover, the fencing distance between subject and the tester during the assessment was normalized and the experimental setup were as in Figures 3 and 4 respectively, which is a standard procedure, following the official rules of the International Wheelchair Fencing Committee.

Although wheelchair fencing technique can be very dynamic, the trunk movements mainly focus on forward and backward in sagittal plane, hence, the analysis was done in two dimensions. In the present study, motions were videotaped by utilizing a Sony 3CCD (DCR-TRV950E) digital video camera recorder and the motion data were further computed by Peak Motus® Motion Measurement System (Peak Performance Technologies). Statistical analysis was run in SPSS version 16.0 (SPSS Inc., Chicago, IL). Due the small simple size, the tested variables were computed by the non-parametric Mann-Whitney method to compare the differences between category A and B, furthermore, the tested variables were 1) Maximum Velocity of Trunk 2) Maximum Angle of Trunk in Lunge and Fast return as Figure 5a and 5b.

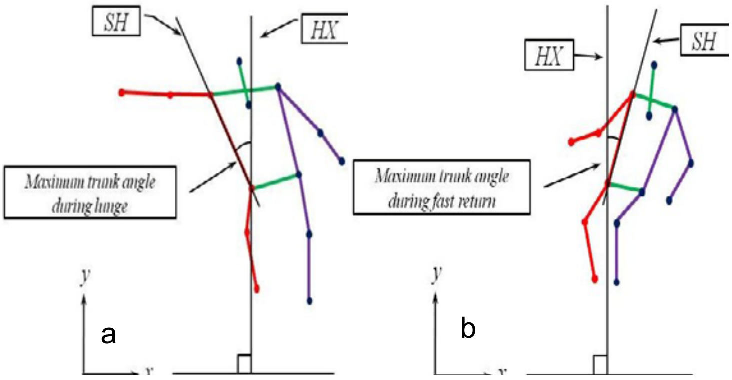

Figure 5: Definition of maximum trunk angle and velocity. In the trials, sterna notch was defined as a reference point to calculate the trunk speed (a) Lunge (b) Fast return. The segmental model used in present study is defined as SH: the line from the shoulder of the fencing arm to the lilac crest of the fencing arm; HX: the vertical line perpendicular to the ground through the hip joint. Furthermore, the angle of experimental result would minus the angle of initial position for normalization.

RESULTS: Table 2 shows the descriptive results, furthermore Table 3 showed that there are no significant differences between category A and B fencers on maximum lunge velocity, lunge angle and fast return velocity, whereas category A fencers could perform significant larger fast return angle in contrast to category B fencers.

Table 2. Descriptive results

	N	Minimum	Maximum	Mean	SD
Cat A Max Lunge Velocity	9	1.04	2.01	1.47	0.28
Cat B Max Lunge Velocity	5	1.30	1.51	1.39	0.08
Cat A Max Lunge Angle	9	33.5	50.5	44.6	5.63
Cat B Max Lunge Angle	5	33.0	46.5	39.8	5.79
Cat A Max Fast Return Velocity	9	0.77	1.52	1.20	0.28
Cat B Max Fast Return Velocity	5	0.91	1.43	1.11	0.22
Cat A Max Fast Return Angle	9	42.0	108.0	64.6	21.43
Cat B Max Fast Return Angle	5	30.0	53.0	40.2	8.58

Remark: The units of Velocity and Angle are in m/s and degree respectively

Table 3. Non-Parametric Mann-Whitney U test

	Max. Lunge Velocity (m/s)	Max. Lunge Angle (degree)	Max. Fast Return Velocity (m/s)	Max. Fast Return Angle (degree)
Z	- 0.535	- 1.535	- 0.601	- 2.469
Asymp. Sig.	0.593	0.125	0.548	0.014*

* $P < 0.05$

DISCUSSION: The purpose of the WFC aimed to classify the fencer's functional abilities with a serial of assessments. Hence, classified participants should have functional differences between categories A and B. In the present study, the maximum trunk velocity and angle in lunge and fast return were assumed as the functional determinants for justifying the outcome of the WFC. However, the results showed that the maximum fast return angle was the only differences between category A and B fencers in the entire tested determinants.

Lunge and fast return are two of the fundamental movements in wheelchair fencing which respect for attack and avoidance of being hit respectively. Hence, the ability to perform a fast and far in this two fundamental movements was critical to wheelchair fencer. When the wheelchair participants perceived this ability, it will encourage the training quality and the competition tactic strategy of victory. Nowadays, WFC is an integrated classification which allows athletes with different disabilities like amputee, polio, cerebral palsy and paraplegics to compete together (IWFC, 2009), so the range that their impairment could affect performance could be very broad. Nevertheless, our results showed that category A and B fencers in the present study could perform similar abilities. The present investigator doubted that it may be due to sports specificity for wheelchair fencing; participants have to perform movement on the wheelchair, so that the lower limb ability will cause negligible limitation for wheelchair fencing whereas trunk control will be a weighty indicator. More and more, although the maximum speed and angle of trunk movement were assumed as an important functional determinant in this study, it didn't represent this is a sensitive indicator to descript the differences between these two categories of fencer, because there is much additional research remains required like sitting balance or the functional agility on wheelchair.

CONCLUSION: This study was the first in the literature to explore trunk ability differences between category A and B fencer in wheelchair fencing, and the result showed that category B fencers could perform a similar trunk performance in most of the tested parameters as category A. This result may provide information to International Wheelchair Fencing Committee for the need of research on WFC to clarify the differences between these two categories of participants.

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A CASE STUDY OF STRIDE FREQUENCY AND SWING TIME IN ELITE ABLE-BODIED SPRINT RUNNING: IMPLICATIONS FOR AMPUTEE DEBATE

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Recent research into trans-tibial double-amputee sprint performance has debated the possible inherent advantages, disadvantages and limitations to sprinting with prosthetic limbs compared to healthy limbs. Biomechanical data gathered throughout a training season from an elite able-bodied sprinter provide a new perspective on this debate. Peak stride frequency was measured at 2.62 Hz, and the corresponding swing time was estimated to be 0.287 s in the able-bodied sprinter. Published swing time and stride frequency values from the double-amputee at maximum velocity, thought to be beyond biological limits, therefore may not be so, although previously published research has provided evidence that some joint kinetic values from the double-amputee have not been shown in elite able-bodied sprinting.

KEY WORDS: athletics, track and field, able-bodied sprinters, double-amputee, limits of maximum performance, kinematics

INTRODUCTION: Current debate on the biomechanical limits, advantages and disadvantages of an elite trans-tibial double-amputee compared to able-bodied maximum performance has reached varying conclusions. A Point: Counterpoint debate by Weyand et al. (in press) highlighted stride frequency and swing time as important limiting factors, but disagreed on the boundaries possible in able-bodied athletes. Stride frequency has previously been identified as an important factor in sprint performance (Bezodis et al., 2008b), although its limits in able-bodied sprinters are currently unknown. The aim of this study was to determine whether double-amputee stride frequency (2.56 Hz) and swing time (0.284 s) values published by Weyand et al. (2009) are within the boundaries possible in able-bodied athletes.

METHODS: One elite able-bodied sprinter (age: 18 years, height: 1.79 m, mass: 81.4 kg, 100 m personal best: 10.20 s) gave written informed consent to participate. The main data were collected in an indoor athletics centre over six months during the ‘speed work’ phases of indoor and outdoor seasons. Two digital cameras (Sony DCR-TRV 900E) were mounted on the wall 6.40 m apart, 4.25 m above track level and 7.20 m from the centre of the lane in which trials took place. Each camera was set up with a shutter speed of 1/600 s and a field of view of 6.2 m in the lane of interest. There was a 2.5 m overlap of the two cameras’ views at the centre of the global field of view. Both cameras were separately calibrated using six control points in the following two orthogonal planes. A calibration area of 6.00 x 1.17 m was set in the transverse plane at track level for the determination of step length. A second calibration area of 5.50 x 2.06 m was set in the sagittal plane at the centre of the lane for the determination of velocity. Data were collected during the course of the subject’s normal training sessions, where the athlete was performing ‘speed work’. Sessions typically comprised six to eight runs in the early spring and three to four runs by late spring and summer. Video images of the runs were recorded during the maximal velocity phase of a sprint. The start of the combined 9.5 m field of view of the two cameras was at least 40 m from the start of the sprint. The subject was allowed normal training recovery after each run.
Video data were imported into Target (Loughborough Innovations Limited, UK) for digitising. The last field before touchdown and the first field after touchdown were digitised for each contact. A 20-point model of the human body was used, with inertia data based on de Leva (1996), apart from the foot segment, for which Winter’s data (2005) were used, with an extra 200 g added to account for the mass of the running spike (Hunter *et al.*, 2004). The toe of the

ground foot was independently digitised three times during the first field after touchdown to minimise error in the calculation of step length. Digitised trial sequences were reconstructed using a 2D DLT routine (Walton, 1981). Calculation of variables for each individual step was always carried out with the data gathered from a single camera, ensuring that the respective calibration was used, i.e. no step variables were calculated from mixed views.
Step lengths were calculated by subtracting the mean of the three reconstructed contact foot toe locations from one contact in the direction of the run from the corresponding mean contact foot toe location of the contralateral foot at the next contact. Step velocity (average centre of mass velocity across the whole step) was calculated as the difference between the mean centre of mass displacements from the two digitised fields at two consecutive contacts divided by the time between them. Step frequency was calculated by dividing the step velocity by the step length. Further details of the calculations can be found in Bezodis *et al.* (2008a). Data were collected from three or four steps within the maximum-velocity phase of each maximal sprint, where a stride is defined as two consecutive steps. Thus, the 141 measured steps could not all readily be grouped into separate strides.
Following the results of the main study, a brief follow-up study was set up outdoors after the competition season had finished. Three-dimensional kinematic data was collected with two high speed cameras (Motion Pro®, HS-1, Redlake, USA) at 200 Hz. Camera 1 was set perpendicularly at 38.30 m away from the midline of the running lane and camera 2 was located at 30.00 m in front of the central point with 1.50 m offset to the right of the running lane in order to let the athlete pass the camera. Both cameras were set at the height of 1.00 m with 1/1000 s shutter speed. The camera views were calibrated with 18 points creating a volume with maximum dimensions of 6.50 m (length), 1.20 m (width) and 2.07 m (height), thus allowing a full running stride to be analysed. The athlete performed three maximum effort sprints with 40 m run-in before the camera views and normal training recovery between the runs. Contact time, swing time and stride frequency data were analysed directly from the camera clips. The stride velocity was analysed in exactly the same way as the step velocity above by digitising the relevant fields from both camera views in Peak Motus® (v. 8.5, Vicon, USA) with centre of mass positional data extracted after DLT reconstruction (Abdel-Aziz and Karara, 1971).

RESULTS: In the main study, the maximum step frequency measured was 5.28 Hz at a velocity of 10.84 $m{\cdot}s^{-1}$ (Figure 1), equivalent to a stride frequency of 2.64 Hz. When considering pairs of consecutive steps the highest stride frequency was 2.62 Hz, at 10.87 $m{\cdot}s^{-1}$. A total of 20 measured steps for this athlete displayed a frequency of at least 5.12 Hz (equivalent to a stride frequency of 2.56 Hz). Comparisons against known locations on the track surface and repeat digitisations in the horizontal plane revealed step length errors of ± 0.01 m. Comparisons of sagittal plane results to sequences in which all fields were digitised revealed velocity errors of ± 0.01 $m{\cdot}s^{-1}$, and errors in step frequency of ± 0.01 Hz. In the follow-up study, the highest stride frequency was 2.53 Hz at 10.60 $m{\cdot}s^{-1}$ (Table 1). The shortest measured swing time was 0.295 s and contact times varied between 0.095 and 0.105 s.

Table 1. Velocity, stride frequency and swing and contact times for three follow-up runs.

		Run 1	Run 2	Run 3
Velocity	[$m{\cdot}s^{-1}$]	10.34	10.61	10.60
Stride frequency	[Hz]	2.44	2.47	2.53
Swing time	[s]	0.305	0.300	0.295
Left contact time	[s]	0.105	0.105	0.105
Right contact time	[s]	0.100	0.095	0.095

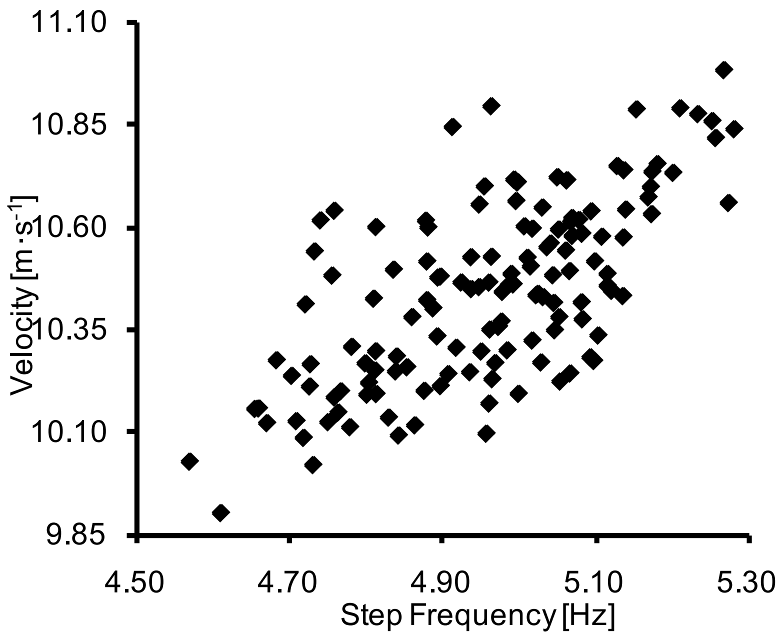

Figure 1. Step frequency against velocity for all measured steps in the main study.

DISCUSSION: Results of the main study showed that 20 of the 141 measured steps in the study were at a greater frequency than the double-amputee strides measured by Weyand et al. (2009). The debate in Weyand et al. (in press) also focused on swing time, which although not directly measured in the main study, can be estimated. The minimum stride time was 0.381 s (reciprocates as 2.62 Hz). An individual stride time comprises ground contact and recovery swing of the same limb. Since contact times were not measured in the main study, they must be inferred from other data. The top speed of the able-bodied athletes in Weyand et al. (2009) was 10.8 ± 0.6 $m{\cdot}s^{-1}$, equivalent to the 10.87 $m{\cdot}s^{-1}$ measured in this study. The subjects in Weyand et al. (2009) had a mean contact time of 0.094 s which corresponds well with the elite athlete (Bezodis et al., 2008b) contact time of 0.097 s measured via force plate in another study (Bezodis et al., 2008a), although the velocity for this athlete on that occasion was slightly lower at 10.37 $m{\cdot}s^{-1}$. These values also match with the shortest contact times of 0.095 s measured from our elite athlete in the follow-up study running at 10.60 $m{\cdot}s^{-1}$. Subtracting 0.094 s from the 0.381 s stride time leaves a 0.287 s swing time for the elite athlete in this study, just longer than 0.284 s recorded for the double-amputee at the same speed (Weyand et al., 2009). Considering that generally able-bodied athletes have slightly longer aerial than contact times (Weyand et al. 2000; 2009), it could be possible that the elite athlete's contact time in this study was slightly shorter than the values presented above. Even if the contact time was 10% less than above, i.e. 0.085 s (which would be extraordinarily short), the swing time would still have been 0.296 s. In the follow-up study during the off-season, the shortest measured swing time was 0.295 s. This occurred with a stride time of 0.395 s (stride frequency of 2.53Hz), i.e. the contact time was 0.100 s at the start of this stride. Overall, it is clear that the contact time as short as 0.085 s in the above example is not realistic for this athlete, and consequently the upper limit estimation for the swing time at 0.296 s is over conservative when the athlete was running with the stride frequency of 2.62 Hz. In any case, these values are substantially shorter than able-bodied athletes' swing times of 0.359 ± 0.019 s in Weyand et al. (2009) and clearly shorter than 0.320 s reported for the fastest 100 m sprinter in the world at the time (Weyand et al., 2000). While the double-amputee's swing time of 0.284 s (Weyand et al. 2009) may still be shorter than any able-bodied sprinter's data, this study shows that swing times under 0.300 s are possible at high speeds for able-bodied athletes and challenge a conclusion in the debate by Weyand et al. (in press) that 0.284 s is beyond biological limits. Moreover, 14% of all steps measured in this study were greater in frequency than the 5.12 Hz measured in the double-amputee sprinter by Weyand et al. (2009), suggesting that a step frequency of 5.12 Hz is clearly possible in an able-bodied sprinter and thus not artificial nor attributable to non-biological factors as claimed in Weyand et al. (in press).

Determining whether an individual biomechanically fits into a population when sample sizes of elite athletes are inevitably small makes consideration of mean and standard deviation prone to incorrect assumptions of normal distribution of data. Presentation of individual able-bodied values for specific variables shows whether the double-amputee produces similar magnitudes to controls. Data presented here suggests the perceived amputee temporal advantages over able-bodied athletes may not exist. However, previous studies of joint kinetics suggest that the double-amputee does clearly get an advantage over able-bodied athletes during the maximum-velocity phase (Bezodis et al., 2008a; Brüggemann et al., 2008). There are so many ways of looking at sprint biomechanics without even considering physiological and psychological factors, as well as ethical and philosophical issues, that it may never be possible to say if there's a definitive overall advantage inherent to amputee performance.

CONCLUSION: Kinematic data presented here show the example of an elite able-bodied athlete with stride frequencies and swing times similar to those recorded in the double-amputee. Data presented elsewhere, however, suggest that some of the joint kinetic variables in the double amputee have not been matched by elite able-bodied sprinters.

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THE ANALYSIS OF GAIT IN INDIVIDUALS WITH TRANSFEMORAL AMPUTATION USING A BIONIC KNEE DESIGN: A CASE STUDY

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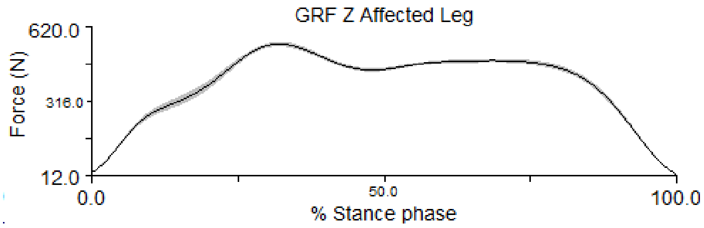
KEYWORDS: biomechanics, gait, amputees, prosthesis, kinetic analysis, kinematic analysis.

INTRODUCTION: Walking is the body´s natural means of moving from one location to another. It is also the most convenient means of traveling short distances (Perry, 1992). Approximately 4500 new amputations of lower extremities are performed annually in the Czech Republic. Worldwide, 200 – 500 million major amputations are performed each year. Major limb amputations of lower extremities account for approximately 85% of all cases of amputations (Ellis, 2005). The loss of a lower extremity causes the inability to walk without a prosthetic aid. Biomechanics has an important place in the multidisciplinary team, which is essential for the complex care of amputees (Janura, Svoboda, Kozakova, & Birgusova, 2006). New prosthetic designs may enable people with transfemoral amputation perform sports (running) activities in the future. However, it is necessary to master walking with a prosthesis first. The aim of our study is to describe the selected biomechanical parameters of gait in patients with a two-year experience with a bionic knee.

METHOD: One female (age – 45, height – 1.62m and weight – 64 kg) with transfemoral amputation participated in the study. The amputation was on the left side, it was executed 30 years ago, and she had been using a bionic knee for 2 years. She was a good walker who used her prosthesis on a regular basis and led a normal active life. The participant visited the laboratory on two separate occasions. On both occasions, the proband performed fifteen attempts to walk across two Kistler force plates (model 9286AA) embedded in the floor. For the purpose of the kinematic analysis, motion measurement markers were placed on the proband's body. We used lower body marker set. Objective gait measurements were acquired with a computerized video motion analysis system utilizing seven infrared cameras (Qualisys). Capture frequency was 247 Hz. Marker data were processed using Visual3D software (C-motion, Rockvile, MD, USA). Using Visual3D, all lower extremity segments were modeled as frustra of cones. The local coordinate system of the thigh, leg and foot was derived from the standing calibration trial. The lower extremity 3-D joint angles were calculated using a Xyz Cardan rotation sequence.

RESULTS:



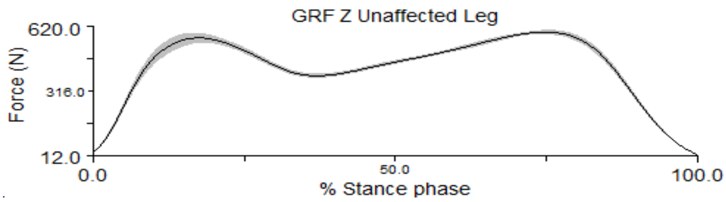


Figure 1. Vertical ground reaction force during the stance phase of the subject with Bionic knee design (n=1, 15 attempts). The gray area displays the standard deviation.

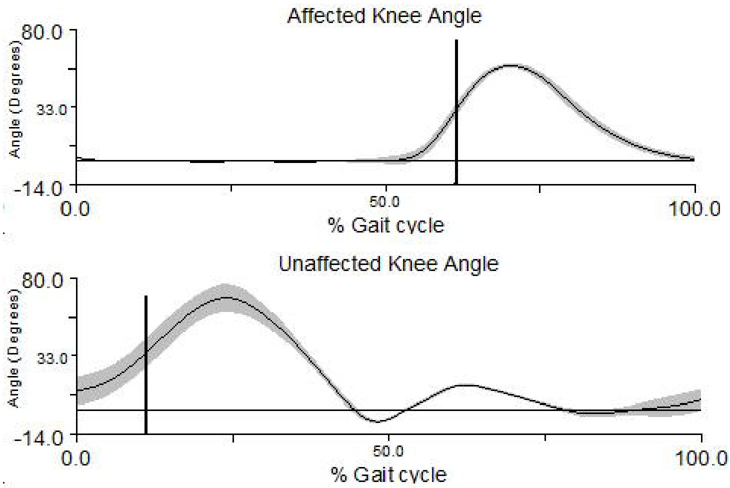


Figure 2. Knee angle during the gait cycle of the subject with Bionic knee design (n=1, 15 attempts). The gray area displays the standard deviation. Each graph begins and ends at initial contact and therefore represents one gait cycle. The vertical line represents toe off for each condition.

The first flexion (Figure 2) typical for normal walk does not occur in the bionic prosthetic knee. This fact also influences the course of the vertical ground reaction force in the healthy extremity in relation to the stance phase. Figure 1 shows that the active propulsion takes place from 30 to 80% of the stance phase on unaffected leg.

DISCUSSION: The purpose of this study was to describe the selected biomechanical parameters of gait in patients with a two-year experience with a bionic knee. Valmassy (1996) states the active propulsion time from 50 to 80% of the stance phase in normal population. Our finding is that the active propulsion during the stance phase on unaffected leg is by almost 20% longer in the stance phase as against normal gait. The phase of active propulsion therefore starts earlier, as Figure 1 shows. The non-existence of flexion in the prosthetic knee joint during the swing phase remains a great problem of walking with transfemoral prosthesis. This fact complicates the possibility of changing the walk to running in patients with a transfemoral amputation.

CONCLUSION: During the gait measurement, there was no flexion during the swing phase in the afflicted extremity of the patient with a two-year experience with a bionic knee joint. This fact also affected the active propulsion during the stance phase of the healthy extremity. These deviations from normal gait complicate changing the walk to running.

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ELECTROMYOGRAPHICAL ANALYSIS OF DOUBLE POLE ERGOMETRY: STANDING VS. SITTING.

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This study assessed the difference in stand-up athlete's muscle activity of the rectus femoris in standing and sitting using a double pole ergometer. Five subjects participated in two technique specific peak VO_2 tests, and a percentage of the maximum scores were used to determine stages for analysis of the electromyography data that was collected. An ANOVA revealed a significant difference in electromyographical activity between ski position and stage using a pre-determined alpha level of $p<0.05$. Separate paired T-tests were used to determine that there was no statistical difference in muscle activity between the two stages in sitting ($t=1.464$, $p=0.217$). However, there was a statistical difference between stages in the standing position ($t= -5.023$, $p=0.007$).

KEYWORDS: cross country skiing, disabled sports, biomechanics, Nordic skiing.

INTRODUCTION: Nordic skiing may date back to 5,000 BC according to cave drawings found in Norway (Clifford, 1992). Initially skis were used as a tool to assist in moving across the snow for practical purposes such as military pursuits (Holmberg, 2003) and hunting (Seiler, 1996). Nordic skiing competition evidence dates back as early as 1520 in Norway (Holmberg, 2003). In 1924 Nordic skiing was included in the Winter Olympic games (Clifford, 1992), and in 1976 was added to the Paralympic Winter games (Nordic Skiing).

One event in Paralympic Nordic Skiing is sit-skiing where the athlete sits on a frame mounted to cross-country skis and employs shortened ski poles and the double pole technique for propulsion over the snow. Double poling is defined when the upper body provides most of the propulsion via bilateral pole pushes (Holmberg, 2003), and is a common technique for both stand-up and sit-skiing.

The double pole technique has greatly developed over the past fifteen years (Lindinger et al., 2009). A number of research articles have been published related to performance and evaluation of the physiological and biomechanical aspects of double poling in the standing position. This previous research has found that double poling requires high levels of upper body fitness, and ski specific upper body fitness has been found to be an important predictor of Nordic ski performance (Mahood et al., 2001). Only one study has investigated double poling in the standing position versus the sit-ski position employed by Paralympic skiers (Tervo & Jensen, 2009). Results revealed that stand-up skiers, using a sit-ski obtained peak velocity at the same point during the poling cycle as a sit-ski athlete. This suggests that it may be possible to use experienced stand-up skiers as subjects for biomechanical sit-ski research, but additional research is needed.

In 2005, Holmberg et al. studied muscle involvement of stand-up skiers using electromyography and found that the rectus femoris (RF) is involved in stand-up double poling. When considering the use of stand-up skiers in sit-ski research, it would be important to know if the RF is activated during sit-ski double poling in this group. Since some sit-ski athletes with paraplegia may have no function of the RF, the use of athletes with intact RF function in sit-ski research could complicate interpretation of results. The current study constitutes foundational research to help determine if stand-up athletes could be used for research of sit-skiing.

METHODS: Five Nordic ski collegiate athletes participated in this double pole study which used two positions: standing and sitting. The mean (± SD) age was 19.8 ± 2.12, height was 174 ± 7.77 cm, and weight was 68 ± 5.8 kg. The current research study was approved through NMU's IRB (#HS09-307). The subjects signed an informed consent and completed a PAR-Q questionnaire (Physical Activity Readiness Questionnaire, Public Health Agency of Canada, 2002) before participating in the study.

Subjects performed two maximal oxygen consumption tests using the double pole technique on a modified VASA Ergometer (Essex Junction, Vermont, USA). A standard VASA Ergometer was mounted at the base of a parallel vertical railing system. The pull cords for poling were directed through pulleys which were mounted to an adjustable cross-bar attached to the railing system, which allowed for height adjustment. In order to keep the double pole as similar to skiing as possible the VASA Ergometer was adjusted according to each skier's classic pole height. The position of the skier's stance on the floor was set at 73 cm horizontal distance from the toes to the VASA for the standing test. The rear edge of the sit-ski seat was positioned 162 cm from the VASA for the sit-ski test (Figure 1). Each subject brought in a pair of their classic poles, and the pull height was set by using the bottom of the cross-bar adjustment at 15% higher than the classic pole height for standing. An adjustable set of poles was used to determine the seated height by using the top of the pole set at the level of their eyebrow when seated in the sit-ski. This height was then measured and the pull height was set at 50% higher than the pole height. The discrepancy of the percentages is due to the fact that static nylon extenders (68 cm) were added to the VASA for the seated test so that the length of the cords were long enough to complete a pole cycle.

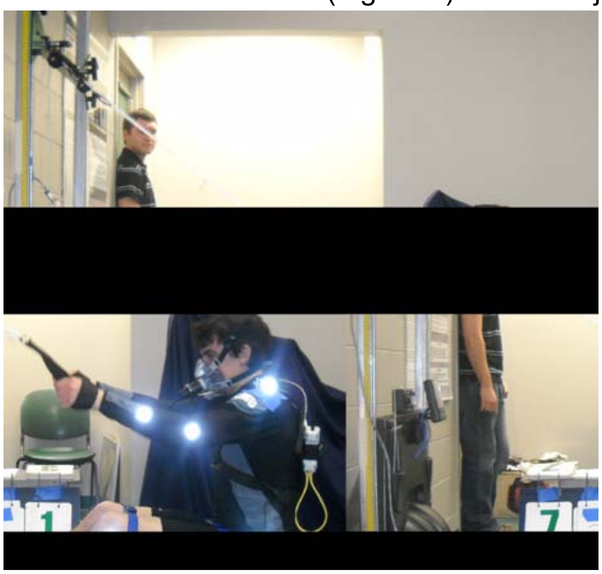

Figure 1. A photo of the VASA Ergometer set-up with a sit-skier.

One maximal test was performed in a standing position to simulate on-snow cross-country skiing. A second maximal test was performed in a seated position on a modified sit-ski to simulate sit-skiing. Both tests were performed at least 24 hours apart. To randomize the order of testing, a counterbalancing method was used. The subject's weight was taken using a Tanita Digital Scale BWB-800A Class III (Tanita Corp., Japan), and a Seca wall stadiometer was used to obtain height to the nearest 0.5 centimeter. The subject was fit with a portable breath-by-breath expired air analysis system to measure maximal oxygen uptake (Oxycon Mobile; CareFusion, CA). The Oxycon Mobile instrumentation weighed approximately 950 grams and was carried via an adjustable vest harness sized to each subject. The instrument flow sensor was calibrated according to manufacturer's auto-cal procedure and the oxygen and carbon dioxide sensors were calibrated with known calibration gases prior to each test. Data for the oxygen uptake were averaged and stored using 60 second averages.

Electromyography data were also collected from the RF of the dominant leg during both maximal tests. A rough abrasion pad was used to abrade the skin, and rubbing alcohol was applied to a 4x4 gauze pad and rubbed over the muscle belly of the rectus femoris in order to take skin oils off of the skin to lessen the impedance seen in the muscle activity. A small drop of Signa Gel (electrode gel Parker Laboratories, Inc.; Fairfield, NJ) was applied to the Norax Dual Electrode (product #272 Noraxon USA; Scottsdale, AZ) before it was placed on the preparation site of the rectus femoris. A ground electrode was placed near the iliac crest on the same side. The

electrodes were connected to the Biopac Systems, Inc. MP 150 (Goleta, CA). Data were processed using AcqKnowledge 3.9.1.6 software, with a gain of 1000, sample rate of 1000, high and low pass filters set at (10Hz) and (500Hz) respectively, and were rectified using root mean square and averaged over 100 ms.
A continuous graded exercise test on the VASA Ergometer was used. The VASA Ergometer uses “an ‘aqua-flow’ flywheel resistance” system with an adjustable damper, and a digital feedback screen which shows cadence, and total time of work-out (Training for Nordic Skiing). A damper setting of 3 was used for all tests and subjects. The protocol included a warm-up and accommodation period of 5 minutes before the test to raise the subject’s VO_2 up to 60-70% of predicted maximum (Howley, 1995). The protocol started with a cadence of 40 strokes per minute which was controlled via an auditory metronome and visual feedback from the VASA Ergometer monitor. The stroke cadence was increased by five strokes per minute each 60 seconds of the test. The test was terminated when the subject chose to terminate due to exhaustion. The data were used in reference to technique specific VO_2 peak.
The oxygen consumption data were averaged and used in order to collect information to determine the subject’s VO_2 at 80%, and 100% of their maximum oxygen uptake. The recorded peak VO_2 score that was closest to the 80% and 100% increments were used to determine the stages of the test.
This stage was documented, and the corresponding EMG data were analyzed using a 2x2 Repeated Measures ANOVA (Position x Stage). A paired T-test was used to assess the differences in technique specific peak VO_2 between sitting and standing positions. All statistical procedures were carried out using SPSS 17.0 (SPSS Inc., Illinois, USA). The predetermined alpha level was set at $p \le 0.05$.

RESULTS: Results of the ANOVA revealed a significant interaction between position and stage (F=65.86, p=0.001). A follow-up paired T-test showed that the difference between seated 80% (70.5 ± 53.1 mV) versus seated 100% (65.4 ± 52.6 mV) was not significant (t=1.464, p=0.217). The difference between standing at 80% (30.8 ± 21.9 mV) versus standing at 100% (56.3 ± 27.6 mV) was statistically different (t=-5.023, p=0.007). A paired T-test demonstrated that peak VO_2 for standing (52.06 ± 3.65 ml/kg/min) was significantly higher than for sitting (38.88 ± 3.53 ml/kg/min), where t=11.16 and p<0.001.

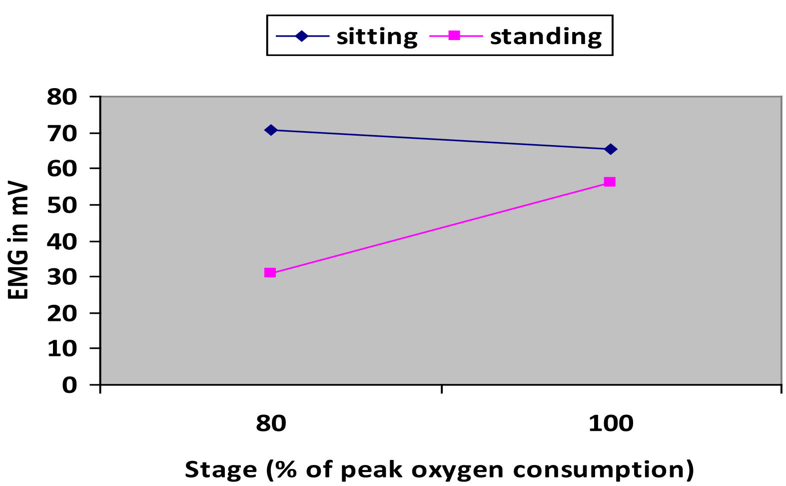

Figure 2. Muscle activity of the rectus femoris response for two stages of a VO_2 max test while standing and sitting

DISCUSSION: The results of this study demonstrate that the RF activity is greater in sitting versus standing. The sit-ski position produced a notable RF activation at both work rates. Thus, RF appears to be highly activated in sit-skiing, or at least more-so than in stand-up skiing. This would indicate a potentially serious limitation to using stand-up skiers for sit-ski research. Since the RF is used to extend the knee and flex the hip, and the seated position naturally places the RF in a shortened position. The shortened position is less optimal for force generation than a more lengthened position (Marginson & Eston, 2001). Theoretically, the length-tension relationship may help to explain why more motor units are recruited for sitting than standing. The RF activation of 80% of max during sit-skiing was similar to that at 100% of max for stand-up double poling, thus we can assume that the RF would be notably activated, if possible, during most competitive sit-skiing intensities. The lack of this activation in skiers with paraplegia would likely require a change in the nature of double poling performance relative to able-bodied skiers. The authors suggest that more research be performed in order to increase the understanding of the biomechanics of sit-ski double pole ergometry.

CONCLUSION: These data suggest it may not be possible to use stand-up athletes for biomechanical sit-ski research, because the stand-up athletes will activate musculature that the sit-ski athletes with paraplegia are unable to activate.

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KINEMATIC PROFILE OF THE ELITE HANDCYCLIST

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KEYWORDS: spinal cord injury, wheelchair sport, paralympics.

INTRODUCTION: A handcycle is a relatively new sports equipment that is a combination of the traditional race wheelchair and a hand operated bicycle crank (Abel, Schneider, Platen, & Struder, 2006). The high mechanical efficiency of this geared fixed-frame racing cycle in comparison to a manual wheelchair can potentially increase the distance a person with a loss of lower limb function can travel. To guide the optimal setup for the handcyclist the influence of crank length (Goosey-Tolfrey, Alfano, & Fowler, 2008; Kramer, Hilker, & Bohm, 2009) and crank configuration (Faupin, Gorce, Meyer, & Thevenon, 2008a; Mossberg, Willman, Topor, Crook, & Patak, 1999) have been investigated. Actual neither research has been done on the upper body kinematics of elite athletes nor on relations between kinematics and performance. The aim of this study was to provide first sport specific information in this area with regards to athletes competing at an international level.

METHOD: Thirteen disabled international level handcyclists (height 176.6 ± 5.9 cm, body mass 69.0 ± 10.35 kg, age 39 ± 9.1 years), who participated in the Paralympics 2008 in Beijing, were recruited for the study. Tests were done on a self constructed, mechanical braked roller system that allowed sport specific testing in the one race handcyle with mounted food rest. Before starting the measurements, the participants completed a 5 min familiarization session to warm up and to get used to the required crank frequency of 90 rpm at approximately 90 Watts. A three-dimensional movement analysis was performed with four Basler Cameras (A602-f, Basler Vision Technologies, Ahrensburg, Germany) operating at 100 Hz and synchronized with a Vicon® MX Unit (Oxford Metrics; Oxford, United Kingdom). The recorded videos were analysed using Vicon® Motus and exported to Vicon® Nexus to determine 3D joint kinematics of the right upper body with the Vicon® upper limb model™. Therefore, thirteen spherical retro-reflective markers were attached to the skin at predetermined anatomic landmarks to define four rigid segments (thorax, upper arm, forearm, hand) with overall seven Degrees of Freedom (DoF). The shoulder joint exhibits three DoF (flexion/ extension, abduction/adduction, internal rotation/external rotation), the elbow joint two DoF (flexion/extension, internal rotation/external rotation) and the wrist joint two DoF (flexion/extension, abduction/adduction), either. The joint angles were calculated using the definition of the Bryant angles. To allow comparison across subjects, the calculated joint angles were normalized to one crank cycle. In the beginning of each session a static trial was acquired to define joint rotation centres. Performance capacity was measured during an additional sport specific stage test. Beijing competition results of each athlete were documented.

RESULTS AND DISCUSSION: The present findings may claim to represent the kinematic aspects of handcycling sport for international elite sports. Since the preparatory work for the determination of force maxima and force minima were collected with internationally active athletes, they can claim a high specificity. On one hand the objective was to adapt sports science research tools to the needs of the relatively new sport of handcycling. On the other hand, the data which were collected with the help of video analyses and their evaluation

were used to analyse relationships between biomechanical factors and performance. Currently there are no studies in the literature that have examined upper body kinematics of international level handcyclists.
The ranges of motion for the shoulder and elbow joints evaluated in this study are, compared to the published values (Faupin, Gorce, Meyer, & Thevenon, 2008b), somewhat lower. These differences may be explained by the examined subjects (elite athletes vs. able bodied persons without handcycling experience). It may be assumed that there was insufficient adapting to the sports equipment for this group within the meaning of an optimum seating position. This would explain that movements were performed, less economical in terms of optimal driving action and that this was followed by greater range of motion. The establishment of angular velocities and angles at certain especially relevant positions should complete range of motion in the scientific debate.
For all analyzed correlations no significant relationship was found between the examined parameters describing shoulder and elbow joint kinematics and the performance parameters (e.g. work load stage test, results Paralympics Beijing). The calculated correlation coefficients partially show a very weak or nonexistent relationship. However, some correlations can be understood at least as an indication of possible favourable configurations. On the one hand, a comparatively large angle of flexion of the shoulder seems to lead to a higher performance in the stage test. On the other hand, it can be seen that a relatively large angle of shoulder abduction was associated with a higher performance during the time trail in Beijing. Against the background of other factors influencing the performance and taking into account the number of participating subjects, these results should not be over-interpreted.

CONCLUSION: This study is a first approach to investigate the kinematic profile of the elite handcycling athlete. The methodology as an adaptation of well reviewed upper limb model provides valid information. No significant relationships were found between the upper limb kinematics and performance. For a clear justification the number of samples should be enlarged.

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AN INDIVIDUALIZED MUSCULOSKELETAL MODEL FOR THE ANALYSIS OF AMPUTEE RUNNING

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The purpose of this study was to develop and apply a three-dimensional full body model for the analysis of transtibial amputee athletes. Sprint running was used as an example with a female sprinter as a subject. Data were collected on a running track leading through a biomechanics laboratory with two force platforms in the runway. Inverse dynamics were calculated using a basic and an advanced model, the latter including detailed information on all important muscle groups. Results support what was published on submaximal running with regard to joint moments and power. The muscle model revealed highly asymmetric muscle forces around the hip joint which may explain the overuse injuries some of these runners experience. Future research is needed to improve the individualisation of the modeling approach.

KEYWORDS: amputee, sprinting, power, muscle activity.

INTRODUCTION: Introduction: Ambulation of unilateral or bilateral amputees has been studied in various papers, partially, to understand neuromuscular strategies adopted by individuals, to understand movement strategies, but also to develop or improve the design of prostheses. Only a limited number of published papers can be identified with detailed mechanical analyses on sports disciplines, such as running or long jump, carried out by amputee athletes (Czerniecki et al., 1992; Nolan et al., 2006). For transtibial amputee runners, it was shown that joint moments change at the knee and hip joint reflecting a greater power generation at the hip but a reduced power flow at the knee. For bilateral amputees similar alterations in power generation across joints were described but obviously the mechanisms in such cases are rather symmetric.
Especially in unilateral amputees, these alterations indicate a remarkably asymmetric pattern of muscle loading which may have implications for training strategies, overuse injuries and, possibly, their prevention.
The aim of this study was to provide a detailed, three-dimensional biomechanical analysis of an elite left-sided transtibial amputee sprinter.

METHOD: One female left-sided transtibial amputee served as subject for the study (syme amputation, level: lower third of left tibia). Full three-dimensional kinematics were collected from the athlete while equipped with 48 reflective markers (similar to Ferdinands, 2004). Markers were placed on anatomical landmarks and defined positions on the prosthesis. Marker trajectories were recorded using a Vicon MX system with eight cameras (250 Hz). Two Bertec force platforms (1000 Hz) were integrated to collect ground reaction forces. The athlete ran five times on a 70 m Mondo running track through the laboratory at submaximal speed (6 m/s).
Marker data were used to calculate joint kinematics, joint forces, moments, and joint powers using a basic customised model (Vicon Body Builder). The mass of the prosthetic was measured beforehand and inertial properties were estimated using basic geometric equations and fluid displacement measures carried out on the testing day.
A second individually scaled model was generated in the AnyBody (AnyBody Tech) modeling environment. In this model the prosthesis was connected visco-elastically to the amputated leg. The model was based on the TLEM model and individually fitted to the changes at the operated leg. The model geometry was individually scaled using an optimisation algorithm in AnyBody. Subsequently, muscle activations were calculated using the optimization

procedure of the AnyBody system resulting in forces generated by all different muscle groups.
After finalization of the set-up and static reference measurements, the subject was given sufficient time to warm up, get used to the force platform and acquire a consistent movement pattern. One difficulty with this was the braking phase after crossing the force plate. A crash mat was provided to assist with which required an extended customization period.

RESULTS: Force characteristics demonstrate a clear impact peak on the right but not on the prosthesis side (Figures 1 & 2). Joint kinematics show relatively small differences while joint moments vary substantially between body sides.

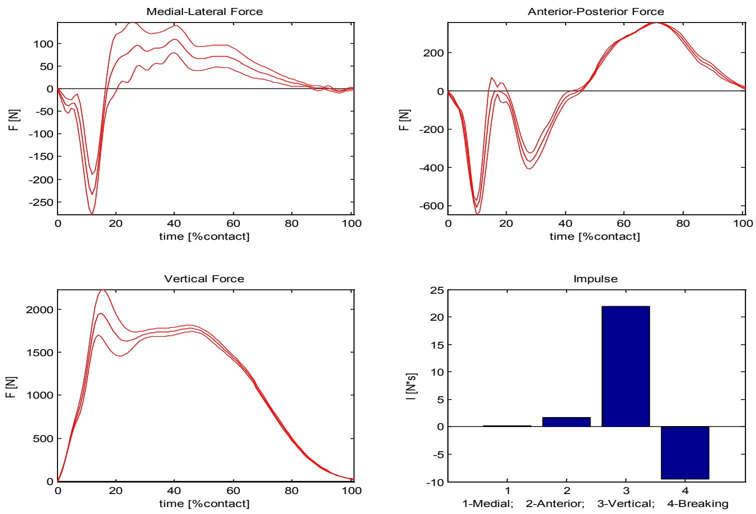

Figure 1. GRF analysis for the right leg at maximum speed (average of three trials).

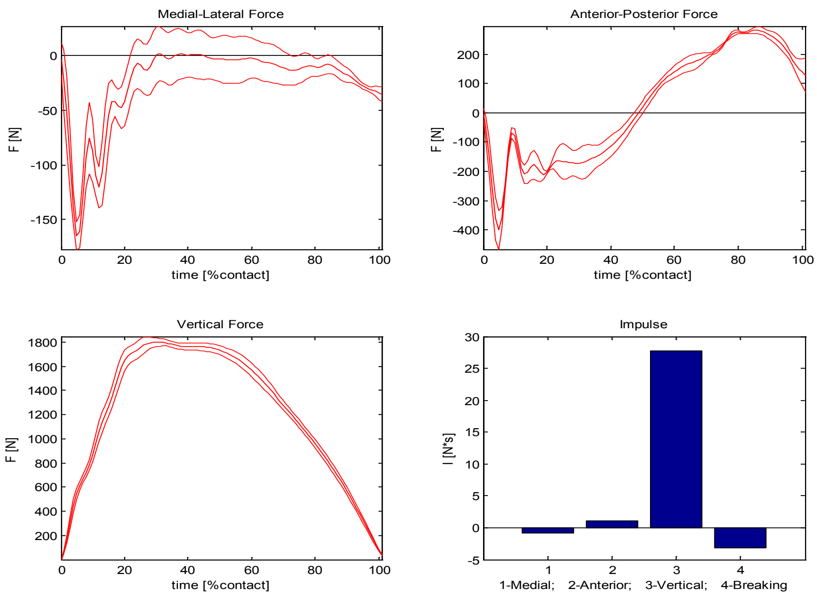

Figure 2. GRF analysis for the left leg at maximum speed (average of three trials).

With regard to the reaction forces, the breaking impulse is greater (i.e., more negative) on the intact side than on the prosthesis side. Still the total positive impulse (2) is larger for the right leg, meaning main propulsion resulting from the right leg. Interestingly, a greater vertical impulse is generated on the prosthesis side at similar contact times. The joint powers were markedly different (Figure 3).

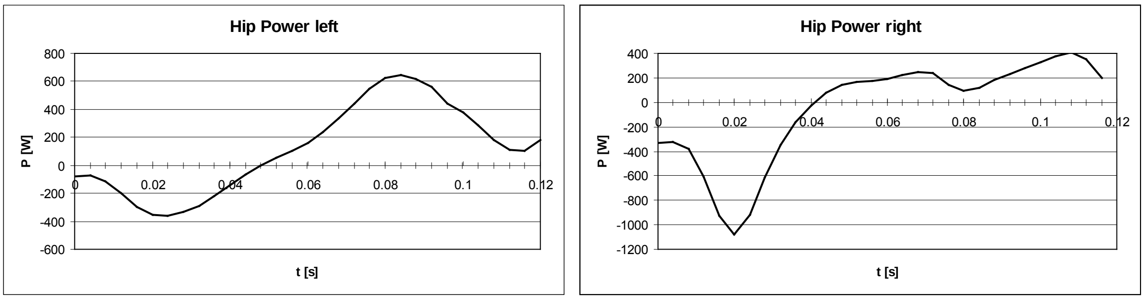

Figure 3. Hip joint power curves from BodyBuilder model.

The centre of mass position with regard to the point of ground contact was compared between legs and showed a highly asymmetric dynamic. This observation matches with the highly asymmetric joint moments at the hip joint and were connected to highly asymmetric muscle forces calculated by the AnyBody model. Substantially higher muscle forces were calculated for both the hip adductors and abductors on the amputated side.

DISCUSSION:

This paper summarises a study of a full three-dimensional analysis of an amputee runner where a detailed individually scaled body model was applied for data analysis. Results from a simple inverse dynamics model are in line with previously published 2D data (Czerniecki et al., 1992) who tested athletes running at 2.8 m/s. However, the differences observed here indicate a more pronounced difference in muscle loading between body sides especially around the hip joint. Additionally, marked changes with respect to trunk and upper extremity movements were observed. These changes reflected by highly asymmetric muscle loads around the hip joint with the main differences found in the ad- and abductor muscle groups. This will have implications for loading on these muscle groups and may help to develop training strategies which may need to be quite different from healthy subjects.

CONCLUSION:

In this study a highly complex model for the analysis of amputee athletes was proposed. Interesting observations were made which match this individuals training induced overloading symptoms. Also, the most remarkable differences were found for movement directions out of the sagittal plane indicating the need for such analyses when aiming at understanding biomechanical mechanisms in amputee running.

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Jumping

VALIDATION OF ACCELEROMETER DATA FOR MEASURING IMPACTS DURING JUMPING AND LANDING TASKS

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The purpose of this study was to examine the validity of a commercially-available accelerometer, as used in the field team sports context. Ten adult participants completed two movement tasks: 1) a drop landing task from 30-cm, 40-cm and 50-cm heights [DLAND], and 2) a countermovement jumping task [CMJ]. Peak acceleration values, both smoothed and unsmoothed, occurring in the longitudinal axis [Y] and calculated to produce vector magnitude values [VM], were compared to peak vertical ground reaction force values [VGRF]. All acceleration measures were moderately correlated (r = 0.45 – 0.70), but also significantly higher than weight-adjusted VGRF, for both tasks. Though the raw acceleration measures were mostly above the acceptable limit for error (> 20%), the smoothed data had reduced error margins by comparison, most of which were well below 20%. These results provide some support for the continued use of accelerometer data, particularly when smoothed, to accurately quantify impacts in the field.

KEYWORDS: validation, accelerometers, jumping, impact.

INTRODUCTION: Accelerometers are commonly used as a tool for injury measurement (Brolinson et al. 2006) and assessment of joint loading (van den Bogert et al. 1999). Increasingly, sports scientists are using accelerometry to assess sporting performance and analyze physical demand, particularly in field team sports (Carling et al. 2009). Of particular interest in this environment is the capacity for accelerometers to provide an objective measurement of impact data (i.e., high-intensity movements involving a rapid change in acceleration), which could then be used to aid in the appropriate planning of recovery and training loads (Duthie et al. 2003). Though their use in this context is rapidly growing, few studies have attempted to validate such devices for quantifying sporting movements.

The current literature provides mixed evidence for the accuracy of accelerometers when measuring impact events. Strong correlations (average $r^2 = 0.812$, $p \leq 0.01$) have been observed between peak ground reaction forces (GRF) and peak accelerations measured at the tibia during a countermovement jumping task (Elvin et al. 2007). Additionally, moderate correlations (r = 0.46 to 0.52, p < 0.001) have been observed between three different accelerometer models and body weight-adjusted force, measured in children during continuous low-intensity jumping and a drop landing task performed from a 23-cm footstool (Garcia et al. 2004). By contrast, other researchers did not observe a linear relationship between uniaxial accelerometer counts and GRF (r = -0.15, p > 0.05), measured during a jumping down task (30.5 cm initial height) (Janz et al. 2003). In addition to the paucity of literature in this area, these studies have limited relevance to the team sports environment.

"Off-the-shelf" devices used in team sports settings often sample data at lower rates (e.g., 100 Hz) than accelerometers used in laboratory settings (3000 Hz) (Zhang et al. 2008). It is not known whether accelerometers sampling at relatively low rates can measure impact events with sufficient accuracy. Though peak impact accelerations are of interest to performance analysts in the field, only one study (Elvin et al. 2007) has examined peak acceleration values (as opposed to acceleration counts or data averaged over time). No studies have placed accelerometers at the base of the neck, or accelerometers that have been integrated with Global Positioning System (GPS) units, both of which are commonly employed in elite sporting environments (Carling et al. 2008). Given these gaps in the literature, the purpose of this study was to examine the validity of a commercially-available GPS-integrated accelerometer, as used in the field team sports context.

METHOD: Ten adults (6 males and 4 females) participated in this study. Participants wore one data-recording triaxial accelerometer, which sampled data at 100 Hz and was embedded within a GPS monitor (SPI Pro, serial no. ASP00725, GPSports Pty Ltd, Australia). The unit was worn in a harness provided by the manufacturer, and orientated such that the Y axis was aligned with the longitudinal axis of the participant. For the criterion measure, vertical ground reaction forces (VGRF) were measured by a portable force plate (model ACG, serial no. 0687, Advanced Mechanical Technologies Inc., USA), sampling at 100 Hz. Post a familiarisation session, participants performed all tasks in one session as follows: (1) drop landing [DLAND] from 30-cm, 40-cm, and 50-cm platforms in a randomized order; and (2) countermovement jumping [CMJ].

Acceleration data for all jumps was downloaded from the monitor using proprietary software (Team AMS version 2.1.05 P2, GPSports Pty Ltd, Australia). A fourth order, zero lag, dual pass, Butterworth digital filter with a cutoff frequency set at 20 Hz was applied to Y axis (Y) and vector magnitude (VM) acceleration histories (Bisseling and Hof 2006). This process was performed in a customised LabView program (National Instruments, USA). Peak acceleration values for both the raw (Y and VM) and smoothed data (Y_s and VM_s) were compared to peak VGRF values, adjusted for body weight (BW). A two-way (measure × task condition) general linear model ANOVA, with Tukey post-hoc test, was used to compare whether the peak acceleration data were significantly different than peak VGRF values. Pearson's r values were calculated to examine the relationship between VGRF and the accelerometer measures. Percent CV difference (%Cv_{diff}), calculated from natural log-transformed data, was used to provide an error measurement; the limit of acceptability was set at 20%. Though it is acknowledged that CV values above 10% are rejected in other fields of research, it has been proposed that this analytical goal is often selected as an arbitrary limit for acceptable variability (Atkinson and Nevill 1998). In consideration of the unconstrained nature of team sports combined with field measurement, 20% was deemed a reasonable and realistic limit of variability for the purposes of this study.

RESULTS: Mean peak Y axis accelerations ranged from 2.24-5.09 *g*, and mean peak vector magnitude values ranged from 2.92-6.04 *g*. Peak weight-adjusted VGRF values ranged from 2.14-4.18 BW. Tukey post-hoc analysis revealed that all peak acceleration values (unsmoothed and smoothed) were significantly higher than VGRF/BW across the tasks. Moderate correlations ($r = 0.45 – 0.70$, $p < 0.05$) were observed between the accelerometer variables and force for all tasks. Most %CV_{diff} values examining unsmoothed data (Y and VM) were above the acceptable limit of 20%. On the other hand, most %CV_{diff} values in relation to smoothed data (Y_s and VM_s) were within the acceptable limit. %CV_{diff} values were lower for CMJ compared to DLAND. Refer to Table I for further details.

DISCUSSION: The use of accelerometers in field team sports is widespread and continues to grow. However, the accuracy of accelerometry for quantifying impact movements in this context is unknown. The results of the present study indicate that, although the raw accelerometer values appear unsuitable as a measure of jumping-based impacts, smoothed accelerometer values can quantify jumping-based impacts with improved accuracy. This is particularly evident in the smaller %CVdiff values (10.9-22.2%), compared to the unsmoothed data (16.8-30.8%). The accuracy of the raw data may have been influenced by monitor placement, and monitor vibration, occurring due to movement within the harness.

Table 1. Accelerometer data compared to vertical ground reaction force data

Task	ANOVA (Main effect of measure)	Pearson r	Percent CV Difference (%)
Comparison to Y axis accelerometer data (Y)			
DLAND	F (4, 745) = 98.0*	0.54*	21.4
CMJ	F (4, 499) = 45.0*	0.49*	16.8
Comparison to vector magnitude (VM) accelerometer data			
DLAND	As above	0.56*	30.8
CMJ		0.45*	22.5
Comparison to Y axis smoothed accelerometer data (Y_S)			
DLAND	As above	0.70*	15.5
CMJ		0.59*	10.9
Comparison to vector magnitude smoothed accelerometer data (VM_S)			
DLAND	As above	0.70*	22.2
CMJ		0.55*	15.9

* Value is statistically significant, $p < 0.05$; DLAND = Drop landing task; CMJ = Countermovement jumping task.

In the existing literature, monitors have been attached to participants at sites within close proximity to the impact site. Both Garcia et al. (2004) and Janz et al. (2003) utilized hip and waist placement sites, as is common practice in physical activity research (Ward et al. 2005). Meanwhile, Elvin et al. (2007) measured tibial axial accelerations by aligning two accelerometers with the fibular heads. It is likely that placement site plays a role in the strength of any linear relationship between impact forces and impact accelerations. It is perhaps unsurprising then, that correlations observed in this study between force and acceleration data were only moderate in strength, given the distance of the monitor (placed at the base of the neck; manufacturer-recommended site) from the impact site at the feet.

It might be expected that accelerations measured at the base of the neck would be lower than the reaction forces experienced on initial ground contact, due to shock attenuation by major body structures (Bennell et al. 1996; Coventry et al. 2006; Lafortune et al. 1996). Previous studies indicate that impact shock experienced at the knee can be attenuated by more than 50% by the time the shockwave passes to the head (Bennell et al. 1996; Coventry et al. 2006). However, the results of the present study showed the opposite outcome, as peak impact accelerations were significantly higher than peak impact VGRF values. This apparently contradictory finding may have several explanations.

The use of accelerometry to measure impacts is based on Newton's Second Law of Motion, which describes a linear relationship between force and acceleration experienced by an object. The data collected for this study was analyzed accordingly, using Pearson's r to examine whether the accelerometer variables correlated with force. However, previous research (Derrick 2004) indicates that there may be dissociation in the theoretically linear relationship between forces and accelerations experienced by the body during impacts, as a result of the segmental nature of human movement. This has been investigated with specific examination of the influence of knee contact angle on the force-acceleration relationship (Derrick 2004; Lafortune et al. 1996). Lafortune and colleagues (1996) observed that, in response to more severe running impacts, knee contact angle increased to improve shock attenuation through the lower limb. This postural change caused a decrease in peak impact forces, but an apparently discordant increase in peak impact accelerations experienced by the legs (Derrick 2004). As it was not measured nor controlled in the present study, it is not known to what extent knee contact angle (and possibly hip angle and upper body movement) may have affected the force-acceleration relationship in this data.

Higher peak accelerations may also be a result of monitor vibration and sensitivity to small shifts in position. Inadequate security of the unit within its pouch may introduce accelerations that are unrelated to the movement events of interest. Further investigation into alternate placement sites and other methods of securing the unit to the athlete is warranted, and may improve the accuracy of the GPS-integrated triaxial accelerometer while ensuring athlete comfort, player safety, and unit accessibility in competition settings.

CONCLUSION: The findings of the present study provide some support for the use of harness-mounted triaxial accelerometers, particularly following data smoothing, to measure jumping impacts, similar to those that occur in field team sports. Further research is recommended into a wider variety of sport-related movements (e.g., running-based impacts), as well as examining the feasibility of different accelerometer placement sites and attachment methods, to minimize monitor vibration.

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RELATIONSHIP OF GROUND AND KNEE JOINT REACTION FORCES IN PLYOMETRIC EXERCISES

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The purpose of the current study was to assess the relationship between peak vertical ground reaction force (GRF) and peak knee joint reaction force (KJRF). Eighteen recreationally active college students performed a countermovement jump, single leg jump, and drop jump from a height equal to their vertical jump. Vertical ground reaction forces were assessed with a force plate and KJRF were assessed using a combination of GRF and video data. A Paired samples t-Test revealed GRF to be significantly greater compared to KJRF for all jumps ($p<0.001$). Regression analysis indicated a linear relationship between GRF and KJRF for all the jumps. The R^2 values for each jump were CMJ=0.990; DJ=0.993; SLJ=0.995. These results indicate that GRF data may be a viable alternative for assessing KJRF.

KEYWORDS: jump landing, impact, knee injury

INTRODUCTION: Plyometric exercises are widely used to assess athletic performance, enhance muscular power (Markovic, 2007), bone mass (Bauer et al., 2001), and to reduce the risk of injury (Hewitt et al., 2001). The impact from plyometric landings has been of special interest because of the association with various knee injuries, including tendinosis, anterior cruciate ligament (ACL) injury, and osteoarthritis (Dufek & Bates, 1991).
Additionally, jump landings have been used to evaluate the intensity of various jumps (Jensen & Ebben, 2007), as well as the potential for an osteogenic stimulus (Ebben et al., 2010). The quantification of the impact force during jump landings, specifically on the knee, is critical for injury prevention and proper exercise progression.
Impact forces on the knee or knee joint reaction forces (KJRF) are calculated by acquiring both kinetic and kinematic data obtained by force platforms and video analysis, respectively (Bauer et al., 2001; Simpson & Kanter, 1997). This method of analysis relies on labor intensive and error prone digitization as well as slow video sampling rates which may miss the critical landing impact (Chappell et al., 2002).
The use of ground and knee joint reaction forces to assess various jump landings has been evaluated (Jensen & Ebben, 2007; Simpson & Kanter, 1997); however, the relationship between these aforementioned variables remains equivocal. Therefore, the purpose of this study is to determine if vertical ground reaction forces are related to knee joint reaction forces, which may allow for an indirect quantification of knee joint forces without the use of video analysis.

METHOD: Eighteen recreationally active college students (Mean ± SD Age = 21.9 ± 3.8 years; Height = 174.2 ± 8.2 cm; Weight = 70.46 ± 13.01 kg) volunteered to participate in the study. Participants signed an informed consent form and completed a Physical Activity Readiness-Questionnaire prior to participating in the study. Approval by the Institutional Review Board was obtained prior to commencing the study. Participants performed no strength training in the 48 hours prior to data collection.
Warm-up prior to the study consisted of three minutes of low intensity work on a cycle ergometer, followed by dynamic stretching including one exercise for each major muscle group. Following the warm-up and dynamic stretching exercises, subjects performed two trials of a standing vertical jump, for maximal height to determine the standardized depth jump box height. Participants rested for five minutes prior to beginning testing. The order of the plyometric exercises was randomly assigned, consisting of three trials of each jump; a drop jump (DJ) from a height equal to the subject's vertical jump height, a countermovement

jump with arm swing (CMJ), and a single leg countermovement jump from the left leg (SLJ). For the DJ, subjects were instructed to drop directly down from the box and immediately perform a jump. Participants were instructed to jump for maximal height in all conditions. Participants rested for one minute between trials.
The plyometric exercises were performed by taking off from and landing on a force platform (OR6-5-2000, AMTI, Watertown, MA, USA). Ground Reaction Force (GRF) data were collected at 1000 Hz, real time displayed and saved with the use of computer software (Net Force 2.0, AMTI, Watertown, MA, USA) for later analysis. Video analysis of the exercises were obtained at 60 Hz from the sagittal view using 1 cm reflective markers placed on the greater trochanter, lateral knee joint line, lateral malleolus, and fifth metatarsal. The left leg was chosen for analysis for modeling consistency between various jumps. Markers were digitized and segment accelerations were calculated using Motus 8.5 (Peak Performance Technologies, Englewood, CO). Acceleration of the joint segment center of mass was determined after data was smoothed using a fourth order Butterworth filter (Winter, 1990).
In order to synchronize kinetic and kinematic data, a signal was used to initialize kinetic data collection which also inserted an audio tone in the video data. Data were then combined into a single file and splined to create a file of equal length at 1000 Hz. Because GRF for all jumps but the SLJ would have been distributed between both feet (and therefore both knees), GRF were divided by two prior to calculation of the KJRF. Knee joint reaction forces were calculated using methods previously used (Bauer et al., 2001; Jensen & Ebben, 2007). Peak GRF and KJRF were defined as the maximum values during the landing phase of the various jumps and were presented relative to body weight.
The data were evaluated for the assumptions of normality (skewness and kurtosis) prior to further analysis (Tabachnick & Fidell, 2007). After screening, the data were compared within jumps (CMJ, DJ, and SLJ) for differences between the peak GRF and peak KJRF using a Paired samples t-Test. A regression analysis was performed to assess the relationship of peak GRF to peak KJRF in each of the jumps. All data were evaluated using SPSS 17.0 (SPSS, Chicago, IL). Alpha was specified as $p = 0.05$.

RESULTS:

Paired t-Tests revealed that the peak GRF was significantly greater than peak KJRF for all jumps ($p < 0.001$) (see Table 1). Regression analysis indicated that a linear relationship was present for peak GRF and peak KJRF for all jumps (see Figure 1).

Table 1. Peak GRF and KJRF (mean ± SD) for the CMJ, DJ, and SLJ (n=18).

	CMJ	DJ	SLJ
GRF (BW)	1.989 ± 0.518*	2.017 ± 0.794*	3.002 ± 0.625*
KJRF (BW)	1.949 ± 0.557	1.943 ± 0.799	2.929 ± 0.628
Percent Difference	2.01%	3.67%	2.43%

* Significantly greater than KJRF ($p < 0.001$)

DISCUSSION: Results indicate that peak GRF values were significantly higher than peak KJRF during the CMJ, DJ, and SLJ. However, regression analysis demonstrated a linear relationship between peak GRF and KJRF. These results indicate that GRF data may be a viable alternative for assessing KJRF.
Jensen and Ebben (2007), employing similar methods for assessing joint reaction forces, found SLJ GRF and KJRF to be significantly higher than CMJ GRF and KJRF. Utilizing different methods, Simpson and Kanter (1997) found KJRF values of 2.6 BW while performing traveling jumps, a type of single leg bound found in modern dance. The KJRF values found during the traveling jumps are similar to that of the SLJ in the present study (3.00 BW). The difference in the KJRF values between the two single leg jumps is likely a function of the movement. The acceleration during landing in the SLJ, and the corresponding force, would be directed more vertically as opposed to the horizontal bounds where the

resultant force would be more horizontal. These studies however, failed to assess the relationship between GRF and KJRF.

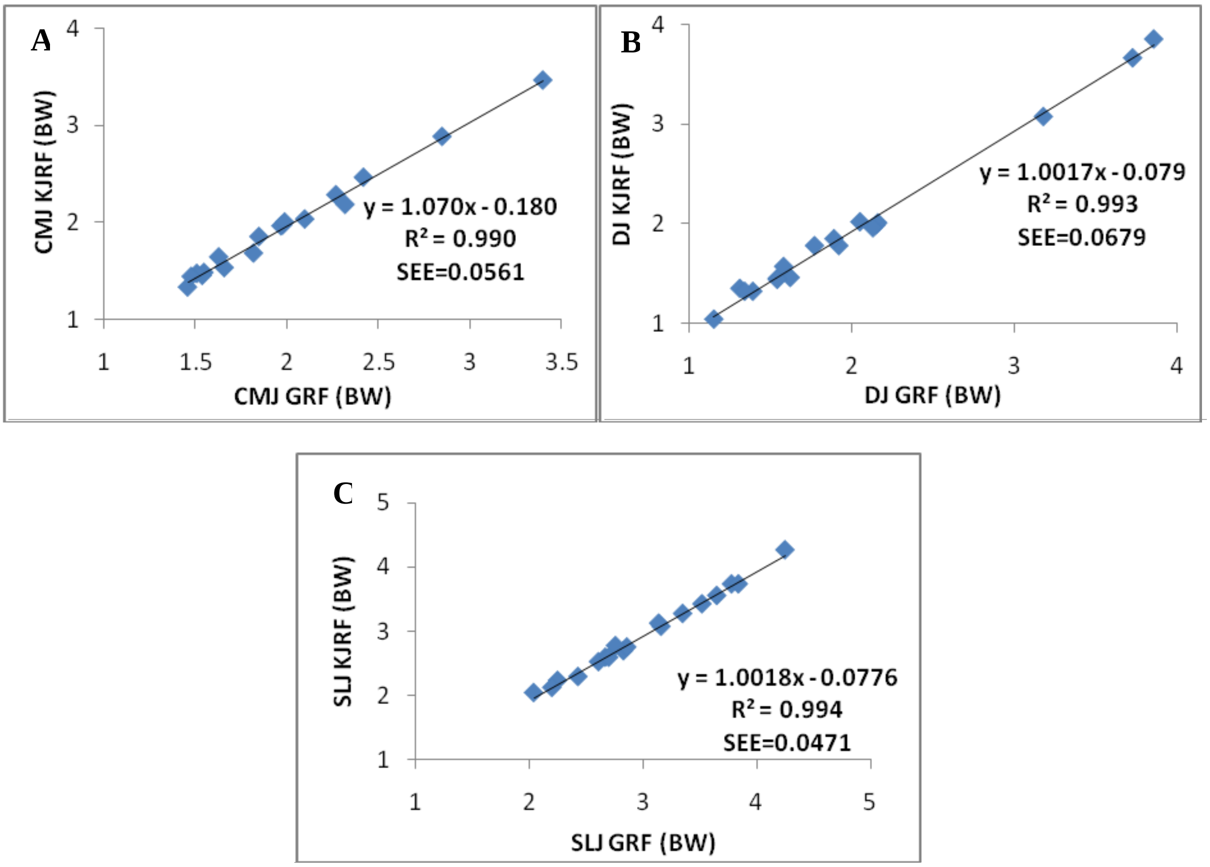

Figure 1. Regression of GRF to KJRF for A (CMJ), B (DJ), C (SLJ) (n = 18).

Previous studies have investigated the correlations between various variables and peak GRF. Results of Peterson and colleagues (2009) demonstrated that peak anterior tibial translation and GRF occurred at the same time when performing drop landings from a height of 40 cm. Anecdotal evidence by Cerulli et al. (2003) indicated that peak ACL strain occurred near the time of peak GRF upon impact during a single left leg jump. Thus, GRF may not only indicate knee joint compressive forces, but also the timing of the tibial translation and ACL strain which are factors that influence ACL injury (Hewitt et al., 2001).
Results of the current study suggest that CMJ, DJ, and SLJ peak KJRF are highly correlated to peak GRF and can be predicted from GRF using linear regression equations. These findings would allow peak KJRF to be calculated without using video analysis, which would be time and labor efficient. However a limitation is that KJRF is not the only variable that should be assessed, as it does not account for shear forces on the knees, which is more important in assessing strain on the ACL (Cerulli et al., 2003). In addition, varus/valgus knee movements are important in assessing the risk of ACL injury (Hewett et al., 2005) and KJRF do not directly account for these variables.

CONCLUSION: Peak GRF values were significantly higher than peak KJRF during the CMJ, DJ, and SLJ, but were highly correlated. These results indicate that GRF data may be a viable alternative for assessing KJRF. The prediction equations for KJRF of the various jumps are as follows: CMJ peak KJRF = 1.07(peak GRF) – 0.18, SLJ peak KJRF = 1.003 (peak GRF) – 0.080, and DJ peak KJRF = 1.002(peak GRF) – 0.079. The ability to assess

KJRF using a force platform will allow practitioners to quantify one of many important variables associated with knee injuries using a relatively inexpensive and time efficient tool.

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THE EFFECTS OF ACUTE WHOLE-BODY VIBRATION ON MAXIMAL COUNTERMOVEMENT VERTICAL JUMP IN RECREATIONALLY ACTIVE MALES AND FEMALES

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KEYWORDS: countermovement vertical jump, whole-body vibration, body-weight squats

INTRODUCTION: Performance is a key factor in any type of training and competition, and even the smallest improvement can have a profound effect on the overall outcome. Some studies have shown that acute whole body vibration elicits a rapid increase in intra-muscular temperature and muscle contraction, thereby enhancing power, strength, and overall performance in the short-term (Cochrane, Stannard, Firth, & Rittweger, 2010). The purpose of this study was to examine the acute effects of three sets of 30 second body weight squats with and without whole-body vibration on maximal countermovement vertical jump in recreationally active males and females.

METHOD: Twenty-six recreationally active males and females, aged 18-23, participated in three days of testing with a minimum of 48 hours between each test-day. Recreationally active was defined as undergoing a minimum of 30 minutes of moderate-vigorous physical activity on three or more days per week. Descriptive demographics are listed in Table 1.

Table 1. Descriptive Demographics

Gender	N	Mean Age	Std.Dev Age	Mean Wt. (kg)	Std. Dev Wt.	Mean Ht. (cm)	Std. Dev. Ht.
Males	14	19.43	.938	75.88	7.67	182.61	6.10
Females	12	20.08	1.311	68.37	9.69	175.86	12.0

Day 1 of testing included baseline measurements consisting of a dynamic warm-up and maximal countermovement vertical jump test after one minute and five minutes of passive rest (standing). The countermovement vertical jump involved a jump and reach technique using a Vertec apparatus in which the individuals were given three trials to reach maximum jump height. The participants were then randomly assigned to the remaining two days of testing. Testing on Days 2 and 3 were identical with the exception that for half of the participants the vibration platform was turned off and for the other half of the participants the vibration platform was turned on. Those participants who completed the testing with the vibration platform turned on during Day 2, completed the testing with the vibration platform turned off during Day 3. Conversely, those participants who completed testing with the vibration turned off on Day 2, completed the testing with the vibration platform turned on during Day 3. All testing with the vibration platform turned on was completed with the vibration platform set at a frequency of 45 Hertz and high amplitude. The sessions on Days 2 and 3 each lasted approximately 15 minutes and adhered to the following protocol: five minutes of dynamic warm-up, two minutes of passive rest (standing), three minutes of training either with or without WBV, one and five minutes of passive rest (standing) each followed by a maximal countermovement jump test. The training on the vibration platform consisted of 3 sets of 30 second body-weight squats. The participants were verbally encouraged to perform the maximal number of repetitions that they could during the 30 seconds for each set. A 30 second rest time was given between each set of body-weight squats, allowing for a 1:1 work to rest ratio.

A repeated measures ANOVA was performed to determine significant differences after the one minute and five minute rest periods. A t-test was performed as a follow-up analysis to examine any differences in jump height between the baseline, vibration and non-vibration conditions at both the one minute and five minute rest periods.

RESULTS: The Wilks' Lambda analysis of the repeated measures ANOVA displayed a significant difference at both the one minute ($p<0.001$) and five minute ($p<0.002$) rest periods. Mean jump height for each condition is as follows: Baseline 1 min: 56.42cm (SD=11.52cm); Vibration 1 min: 58.13cm (SD=12.45cm); Non-vibration 1 min: 58.52cm (SD=11.64cm); Baseline 5 min: 56.61cm (SD=11.36cm); Vibration 5 min: 58.32cm (SD=11.49cm); and Non-vibration 5 min: 57.83cm (SD=11.66cm). Results for the t-tests between conditions are listed in Table 2.

Table 2. T-test Results for vertical jump height in cm

Effect	Mean Diff.	Std. Dev	t	Sig. (2-tailed) .	Effect Size
Baseline 1min- Vibration 1 min	-1.7096	2.8724	-3.035	.006	.523
Baseline 1 min- Non-vibration 1 min	-2.0992	2.4851	-4.306	<.0001	.523
Vibration 1 min- Non-vibration 1 min	-.3896	2.4823	-.800	.431	.523
Baseline 5 min- Vibration 5 min	-1.7096	2.1535	-4.048	<.0001	.512
Baseline 5 min- Non-vibration 5 min	-1.2212	2.4225	-2.570	.017	.512
Vibration 5 min- Non-vibration 5 min	.4885	1.7626	1.413	.170	.512

DISCUSSION: Results of this study indicate that conducting body-weight squats, either with or without WBV, prior to performing a maximal power movement does have a significant improvement on vertical jump after both one minute and five minutes of passive rest. However, no significant differences were observed in jump height after body-weight squats with vibration compared to body-weight squats without vibration after either the one minute or five minute passive rest periods. Therefore, the motion of the squat may be a notable factor in performance rather than the addition of vibration to a training program or competition.

CONCLUSION: Although there were significant differences between baseline jump heights and body-weight squats both with and without WBV, no significant differences were observed between the vibration and non-vibration jump heights following either the one minute or five minute rest periods. This indicates that WBV may not provide enough muscle stimulation to have an impact on jump height, as previously suggested. Therefore, future research is needed to identify how WBV may be effective in improving jump performance.

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ISBS 2010

Oral Session 02

Weightlifting

PRELIMINARY STUDY: INTERPRETATION OF BARBELL BACK SQUAT KINEMATICS USING PRINCIPAL COMPONENT ANALYSIS

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The purpose of this study was to reduce the number of kinematic variables of the barbell back squat for easier interpretation by coaches and athletes. Young active adults (*N*=25) performed the back squat with an intensity of 60%. A total of 10 lower body and trunk measurements were considered for principal components analysis (PCA). Based on the PCA, two components were revealed. The primary component related range of motions (ROMs) in the ankle and knee joints with greater peak flexion angles of ankle, knee, and shank and thigh segments. A secondary component related hip ROMs and hip posterior displacement with greater hip and trunk segment peak flexion angles. Based on this analysis, coaches teaching the barbell back squat should consider two sources of movement variability, one above and one below the hip.

KEYWORDS: squat, kinematics, principal component analysis.

INTRODUCTION: The barbell back squat is a popular exercise among athletes and individuals who participate in resistance training. The National Strength & Conditioning Association (NSCA) initially published a position statement to guide coaches on how to instruct lifters to perform the squat correctly and safely based on a review of the available biomechanical research (Chandler & Stone, 1991). These guidelines have become a popular instruction manual among strength and conditioning coaches and personal trainers for two decades. More recent biomechanical studies have further measured joint kinetics and kinematics of the squat, from both performance and clinical perspectives (Flanagan & Salem, 2007; Fry, Smith, & Schilling, 2003; Salem, Salinas, & Harding, 2003). These studies have reported joint range of motions (ROMs) (Kongsgaard, Aagaard, Roikjaer, Olsen, Jensen, Langberg et al., 2006), peak flexion angles of the lower extremity joints and segments (Flanagan & Salem, 2007; Fry et al., 2003; Salem et al., 2003), and displacements of hip and barbell (Donnelly, Berg, & Fiske, 2006) to evaluate the lifting technique. The NSCA position paper (1991) also cites that injuries to the low back and knee during squatting are common among athletes who are undertrained or have poor technique, further emphasizing the importance of the biomechanical research on proper squat mechanics.

While the variables measured in biomechanical studies show important aspects of the squat kinematics, the sheer number of variables reported may be overwhelming for coaches when qualitatively analyzing a lifter's squat technique. There is a need to reduce the numbers of the kinematic variables to focus on fewer components to evaluate the squat performance.

Therefore, the purpose of the current study was to reduce the number of kinematic variables of the barbell back squat using a principal components analysis (PCA) to summarize the inter-correlated variables into components and create fewer variables to explain and analyze the kinematics of the back squat. A similar design has also been used in gait studies, and the PCA is a commonly used statistical procedure to reduce the number of variables for easier interpretations and analyses (Chester & Wrigley, 2008). Even though the PCA does not provide statistical significance, its information may benefit coaches and athletes to narrow the view points when assessing the squat technique.

METHODS: Twenty-five active college-aged students (20 male, 5 female) volunteered for this study (21±4 yrs.; 179±8 cm; 83±13 kg). All participants were relatively experienced in resistance training including the barbell back squat and free of injuries at the time of data collection. Those who were unfamiliar with the task were eliminated during the recruitment procedure to minimize measurement variance. A university Institutional Review Board approved all procedures and all participants provided their consent before testing.

All participants reported to the laboratory for data collection, and procedures of the testing protocol were provided. Each participant had an adequate amount of stretching and warm-up to replicate a regular training session. In order to normalize the footwear condition among participants, each wore weightlifting shoes (WerkSan, USA) to perform the barbell back squat. A 60-Hz Panasonic digital camera (Osaka, Japan) was placed approximately 1.3 m high and 5 m away on the left side to capture two-dimensional back squat kinematics in the sagittal plane. Reflective markers were placed on left side of the participant's 5^{th} metatarsal joint (toe), lateral malleolus (ankle), lateral femoral epicondyle (knee), and greater trochanter (hip). An additional marker was placed on the end of the barbell. These five markers were used to create trunk, thigh, shank, and foot segments to calculate joint and segmental angles. A segment from the hip to the end of the barbell was used to approximate the trunk segment since the end of the barbell is in the fixed position of the shoulder joint (Fry et al., 2003; McLaughlin, Dillman, & Lardner, 1977).

As all participants were familiar with the barbell back squat, just a brief instruction was given to ensure the left foot was perpendicular to the camera position and the feet were pointed forward for proper tracking of the squat motion in the sagittal plane. If a participant felt uncomfortable with his or her feet pointing forward, practice sets were provided. Also, for those who were not familiar with the weightlifting shoes, practice sets were offered to gain familiarity with the shoes. In order to achieve a comparable effort level from all participants, all trials were performed at 60% of 1RM. For each squat repetition, subjects began standing erect with the barbell on the upper back and descended until the thigh segment was roughly parallel to the floor, and then ascended back to the starting position. Each participant performed a set of five repetitions. The squat video was captured and the data were directly imported into Vicon Motus version 9.2 software (Centennial, CO) for motion analysis.

Two of the five repetitions were averaged and used for calculation purposes. The two repetitions chosen were the third and fourth repetitions in all participants. The kinematic variables measured are shown in Figure 1: (a) trunk segment angle, (b) hip joint angle, (c) thigh segmental angle, (d) knee joint angle, (e) shank segmental angle, and (f) ankle joint angle. ROMs, peak flexion angles of hip, knee, and ankle joints, and angles at the maximum descent position of the trunk, thigh, and shank segments were considered. In addition, a posterior hip displacement was also included. The 10 variables of the squat kinematics were subjected to PCA using Statistical Package for Social Sciences (SPSS) version 17 (Chicago, IL). The PCA was chosen for specifically to reduce the number of the dependent variables by grouping those that are highly correlated with one another.

RESULTS: Prior to performing the PCA, the appropriateness of data was assessed. Inspection of the correlation matrix revealed the presence of many coefficients of 0.5 and above. The Kaiser-Meyer-Oklin value was 0.65 (exceeding the recommended value of 0.60), and Bartlett's test of Sphericity reached statistical significance (p<0.001), indicating the factorability of the correlation matrix. The PCA revealed the presence of 2 components with eigenvalues exceeding 2, explaining a total of 77.3% of the variance (52.7% and 24.7%, respectively). This was supported by the results of the parallel analysis, which showed only two components with eigenvalues exceeding the corresponding criteria values for a randomly generated data matrix of the same size. In order to better interpret the two components, oblimin rotation was performed. The rotated solution revealed the presence of a simple structure with both components showing a number of strong loadings and all variables loading substantially on only one component (see Table 1).

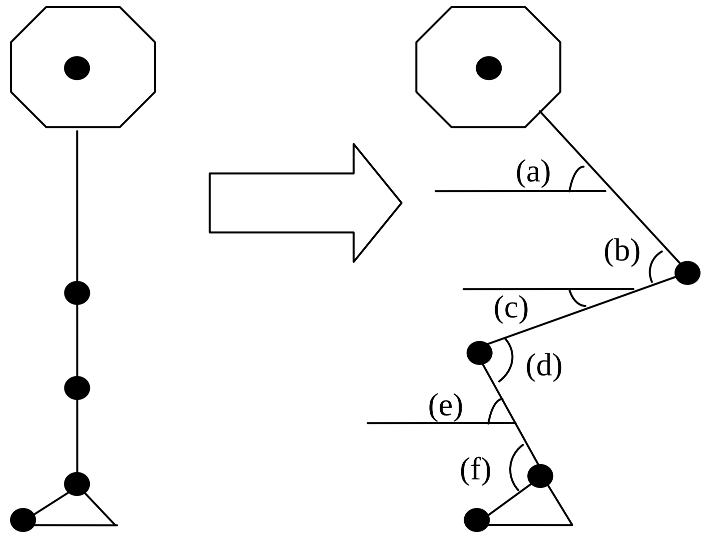

Figure 1. Diagram of kinematic variables measured during squat.

Table 1. Pattern Matrix

Variables	Component 1	Component 2
Knee peak flexion	0.929	
Ankle peak dorsiflexion	0.902	
Shank peak flexion	0.897	
Knee ROM	-0.884	
Ankle ROM	-0.877	
Thigh peak flexion	0.740	
Hip peak flexion		-0.780
Hip posterior displacement		0.742
Hip ROM		0.728
Trunk peak flexion		-0.699

DISCUSSION: The purpose of the current study was to reduce the large number of kinematic variables of the barbell back squat to summarize the inter-correlated variables into components and create new and fewer variables to explain the kinematics of the back squat. The results revealed two independent components. The first component seems to gather the variables from below the hip. It can be interpreted that greater ROMs in the ankle and knee joints created greater peak flexion angles of ankle and knee joints and thigh and shank segments. The second component involved kinematic variables at or above the hip. Based on the pattern matrix in Table 1, it can be interpreted that greater hip ROM leads to greater hip and trunk peak flexion angles and hip posterior displacement. This component seems accurate from a practical view that a greater amount of hip flexion leads to a shift of the pelvis in the posterior direction, and as a result, the lifter leans forward more to balance out the body position at the peak descent position of the back squat. This outcome is also consistent with Fry et al. (2003), who

investigated kinematic differences on restricted and unrestricted knee position during the back squat. Their results showed that the restricted knee position (i.e. knees stay over the toes) caused greater hip and trunk flexion angles to lower the barbell to a desired height as compared to the unrestricted knee position.
Another interesting part from this analysis is that two components were separated clearly from above and below hip, indicating that peak hip and trunk flexion angles are independent of ankle and knee kinematics. Coaches and athletes can use this information to focus on two distinct areas when analyzing the squat performance. For example, if a lifter exhibits excessive trunk flexion, it is most likely not caused by the lower extremity kinematics, but rather movements at the hip with greater flexion and posterior displacement. Therefore, correcting excessive trunk flexion through changes in knee and ankle flexibility may not be the best solution. Another example is that a lack of hip flexibility is very common among athletes (Brophy, Chiaia, Maschi, Dodson, Oh, Lyman, et al., 2009), but it does not necessarily reflect ankle and knee ROM limitations during the back squat. To compare results of the present study with an outcome from Dollenney et al. (2006), the downward gaze of lifters relates to the movement of the trunk and hip, but may not relate to the lower extremity kinematics.

CONCLUSION: The PCA revealed two main components that affect squat technique: one involving movement at or above the hip and one involving movement below the hip. These two components are independent of one another. To evaluate squat performance, coaches and athletes can qualitatively analyze a lifter's technique by simply focusing on these two components.

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INERTIAL SENSOR FEEDBACK DURING SQUAT EXERCISE

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The half-squat is the most widely used exercise in the resistance training, which must be considered optimal only if it is specific and safe. Safeness relies, with other factors, in using the correct technique and being provided with adequate monitoring and feedback. In this perspective, this study a) provided a thorough characterization of the less dangerous squat technique, and b) showed how wearable inertial measurement units (IMU) can be used to quantify key variables useful to reduce errors. The IMU estimate presented a good concurrent validity (r=0.91) for trunk maximal forward inclination, although with significant mean systematic bias of 7±5 deg, and fair concurrent validity for pelvis and barbell rotations in the frontal plane with lower systematic biases. Thus the use IMUs to provide practitioners a quantitative feedback of the execution is encouraged.

KEY WORDS: inertial measurement unit, monitoring, half-squat technique

INTRODUCTION: The so-called half-squat is probably the most used training mean in the development of lower limbs muscular efficiency during physical activity and sports. Despite its undisputed efficacy in this respect, a training program based on overloads can not be considered optimal if it is not, at the same time, specific and safe. In the authors' opinion, safeness relies on the combination of several key elements: a prepared, wise, and careful trainer, a grown and evolved athlete, a controlled and correct technical execution and an appropriate assistance. In particular, technique, certainly along with other factors, may determine differences in the level of risk involved (Aaberg 2000).
Different body structures are involved in this multi-joint exercise, the vertebral column being the most overloaded and thus, most at risk. A slight forward lean of the trunk, due to hip flexion, paralleled by trunk extensor moments, characterizes the half-squat (Wretenberg et al. 1996). This lean should be controlled so that stress on the lumbar spine is kept to a minimum (Neitzel, Davies 2000). In fact, it is known that trunk bending is the factor that most influences the compression load on the lumbar area during execution of the half squat (Cappozzo et al., 1985) and that compressive (Braidot et al., 2007; Zatsiorky and Kraemer 2008) and shear forces (Braidot et al., 2007) as well as intradiscal pressure (Adams and Dolan, 1995) increase for increasing inclinations of the trunk thus amplifying the risk of injury. Nevertheless, papers describing possible "correct" techniques for the trunk movement (Braidot et al. 2007; Kritz et al. 2009) paid limited attention to the assessment of relative risks. Aside from flexion-extension, torsional and lateral bending movements may as well damage of the spine (Adams & Dolan, 1995). Crucial, in this respect, is barbell control. Its asymmetric placement or a poor control of the free weights may lead to swings from side to side as the person tries to lift them, and to a malalignment of the entire column, resulting in the application of considerable and not well-controlled lateral bending forces to the spine.
Thus, to maximize safety, due importance must be given to monitoring that the correct technique is carried out. A measure of movement-related data could contribute in monitoring the exercise and as an aid in correcting errors, as far as the relevant setup is kept simple and allows the subject to perform his/her activity in the real setting. In this respect, wearable inertial measurement units (IMU), that provide accelerations and angular velocities, seem a proper instrument for safety monitoring, while being able to provide information on performance related aspects such as strength and power. Although in recent years the availability of wearable IMUs opened new perspectives in sport sciences, no study, to the authors' knowledge, considered them in this perspective.
A better description of the technique is necessary and central to the development of any technology-based feedback system. Such aid could be fully exploited only if well rooted in a

preliminary qualitative analysis guiding the identification of the most appropriate quantitative parameters. The purpose of this study is, to verify the feasibility of using IMUs to provide feedback to practitioners performing the squat exercise through the following steps: a) characterize the techniques less dangerous and as complete as possible, b) identify common errors, potentially dangerous, c) identify parameters, measurable using inertial sensors, that can be used by practitioners to monitor and, possibly, reduce execution errors, and d) test the reliability of these parameters against reference measures.

METHODS: Characterization of half-squat technique. The following technical description was determined, based on a careful and exhaustive analysis of the literature (O'Shea 1985; Aaberg 2000; Escamilla 2001; Braidot et al. 2007; Comfort & Kasim 2007; Zatsiorsky & Kraemer 2008; Kritz et al. 2009 , Paoli et al. 2009), and used in the experimentation:

Preparatory position: slight extension at the thoracic spine level, scapulae adducted, slightly lift the chest, without forcing the flattening of the lumbar spine.
Barbell: on trapezius and rear deltoids just below C7.
Grip: slightly wider than the shoulders, thumbs blocking the fingers, wrists in slight dorsal extension.
Head and neck: in natural position, gaze straight.
Thoracic spine: slightly extended.
Lumbar spine: natural and stable position, avoiding excessive flexion/extension.
Increase intra-abdominal pressure: abdominal muscle contraction, breath control (before - forced inspiration, descent phase - breath-held, end of the ascending phase - out) and use weight lifter's support belt.
Pelvis: stable, avoid excessive tilt.
Knees: do not exceed 80-90 deg flexion, stable, avoid lateral-medial movements, restrict the antero-posterior ones, maintain knee and toes in alignment.
Feet: flat and stable, width slightly greater than that of the pelvis, externally rotated by 20-30 deg, heels in contact with the ground at all times.

Experimental study: Twelve male subjects (24±2 years, 73±7 kg, 1.79±0.06m), not sedentary, with previous knowledge of the task, and without clinically significant injuries at the most stressed joints, volunteered for this study, after signing an informed consent. The subjects were asked to perform the described technique, wearing a weightlifting belt during a test, for the determination of their maximum load (1RM, 5-6 tests, max 2 reps) and an incremental load test (set at 20,40,60,80% 1RM, repetitions performed for 8,6,4,2 times, respectively). 3D acceleration and angular velocities of the trunk were acquired at 100Hz by a wearable inertial sensor equipped with an on board data logger (Freesense, Sensorize, Italy), fixed onto the weightlifting belt (L2-L4 level). Barbell, trunk, and pelvis kinematics were measured acquiring at 100 Hz the kinematics of the spinous processes of C7, S1, anterior and posterior superior iliac spines, and the extremities of the barbell with a nine cameras stereophotogrammetric system (Vicon MX, Oxford, UK), Figure 1.

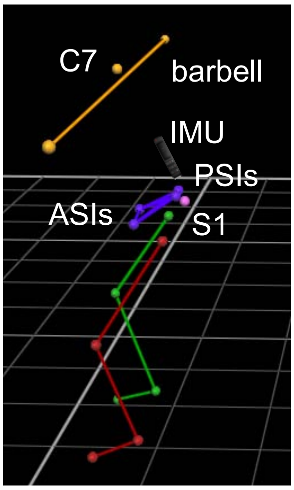

Figure 1: Marker and IMU placement.

Data analysis: Data analysis was performed using Matlab software (MathWorks®, USA). The movement of barbell, pelvis, and trunk, assumed to be a rigid segment, were analysed in the sagittal, frontal, and coronal plane. Acceleration and angular velocity measures, provided by the IMU with respect to a moving reference frame, were independently determined and analysed. When the sensor's inertial acceleration was close to zero, the accelerometer measured the inclination of the sensor relative to gravity. Conversely, when the sensor underwent a motion that generated inertial accelerations, the orientation angles were estimated integrating the angular velocity signal provided by the gyroscopes. A quaternion based algorithm (Favre et al., 2006) and a Kalman filter were implemented in order to compute trunk orientation angles relative to a global reference frame (pitch → sagittal plane, roll → frontal plane). Yaw (→ transverse plane) was provided with respect to the posture. Qualitative analysis for potentially dangerous error identification was performed by visual inspection of the 3D reconstruction of marker trajectories. On the sagittal, frontal, and coronal planes, the rotations and their agreement with the proposed technique were observed. Although the kinematic analysis could be performed with more markers on various technical elements related to the proposed technique, results will be given only for the errors for which it is hypothesised that an inertial sensor on the belt could provide a feedback (barbell, trunk, and pelvis rotation). Each qualitative error was associated to an angle for both kinematics and inertial sensor data, Table 1. For all angles listed in Table 1 the maximal peak for each repetition was used as a quantitative parameter. Trunk orientation in the sagittal plane was assessed at the beginning (minimum) and the end (maximum) of the descending phase.

Table 1 Qualitative description of the main errors and description of the relevant angles selected for IMU and stereophotogrammetric quantification.

Segments	*Qualitative description*	*IMU*	*Stereophotogrammetry*
Barbell	rotation on the frontal plane	roll	inclination with respect to the horizontal plane
	rotation on the transverse plane	yaw	rotation with respect to the reference medio-lateral axis
Trunk	tilt on the sagittal plane	pitch	C7-S1 vector: tilt on the sagittal plane (0° when vertical)
Pelvis	frontal lifting/drop	roll	midPSIs-midASIs vector: inclination on the frontal plane
	rotation on the transverse plane	yaw	midPSIs-midASIs vector: rotation in the transverse plane

The reliability of the parameters provided by the IMU was analysed against reference measures (stereophotogrammetry) with the following statistical analysis, performed using SPSS (version 17.0): 1) Descriptive statistics (Mean ± standard deviation); 2) Error (Stereo-IMU) description through mean bias; 3) Association between reference and IMU as the Mean ± standard deviation of the correlation coefficients of each subject (Pearson); 4) Test of normal distribution of the error with the Shapiro-Wilk test ($p>0.05$); 5) Presence of a linear trend between the amount of random error and the measured values (heteroscedasticity) investigated through inspection of Bland–Altman plots and correlation analysis; 6) If non normal or heteroscedastic, data were logarithmic (natural) transformed prior to agreement statistics; 7) Differences, between IMU and reference values and the effect of the load were ascertained by means of a 2-way fully repeated ANOVA.: 4 (load) x 2 (methods); 8) When load had no significant effect on variables, absolute reliability for repeated measurement was assessed in terms of Limits of Agreement (Bland & Altman, 2007).

Table 2 Descriptive statistics (Mean ± sd) for the two methods and for their difference and agreement analysis (Correlation coefficients and Limits of Agreement).

	sagittal plane		*frontal plane*		*transverse plane*	
Method	*Trunk (bent)*	*Trunk (standing)*	*Barbell*	*Pelvis*	*Barbell*	*Pelvis*
Stereo	36±4 deg	17±1 deg	3±1 deg	9±3 deg	3±1 deg	14±5 deg
IMU	29±7 deg	4±4 deg	4±2 deg		3±2 deg	
Stereo-IMU	7±5 deg	13±4 deg	-2±2 deg	5±3 deg	-1±2 deg	10±4 deg
Correlation	0.91±0.11	0.67±0.37	0.63±0.37	0.75±0.34	0.43±0.26	0.40±0.31
Limits of Agreement	-3.7÷17.3	-21.7÷-3.6	-1.5÷5.5	-	-3.3÷4.7	-19.3÷-1.5

RESULTS: Descriptive statistics of the measures and of the error and correlation analysis are given in Table 2. Since most of the data presented non normal distribution and heteroscedasticity was revealed for maximal trunk inclination, data were log transformed. All parameters, except for barbell rotation in the transverse plane, differed significantly ($p<0.01$), when measured with the two systems. Pelvic rotations on the frontal plane were dependent from load ($p=0.028$).

DISCUSSION: The agreement of IMU and stereophotogrammetry in determining key trunk and pelvis angles during squat was evaluated. The IMU estimate presented a good concurrent validity ($r=0.91$) for trunk maximal forward inclination, although with significant mean systematic bias of 7±5 deg. Fair concurrent validity was shown for pelvis and barbell rotations in the frontal plane with lower systematic biases. In both cases, bias can mainly be attributed to the different portion of the body whose rotation is being measured. For increasing trunk inclination, the difference between systems decrease; this effect may be explained considering that the inclination of the C7-S1 segment, assumed to be rigid with the trunk, can increase due to an increased kyphosis that, in turn, entails a decrease in lumbar lordosis, which reduces the inclination of the L2-L4 portion and, thus, of the IMU.

CONCLUSION: The half-squat, so widely used, requires careful monitoring and extreme competence of coaches and athletes. The technique proposed, which aims to reduce the risks of the exercise, is therefore the primary result. The validity of the IMU estimate of maximal trunk inclination encourages in using such device to provide the athlete and his/her coach a quantitative feedback of the execution; helping them to improve the technique and potentially increasing their awareness about the risks related to this dangerous exercise.

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THE EFFECTS OF WEIGHTLIFTING SHOES ON SQUAT KINEMATICS

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Athletes may not always consider footwear when performing the barbell back squat during training. Several footwear companies have designed shoes claimed to enhance performance in weightlifting and powerlifting. The purpose of this study was to compare the kinematics of the barbell back squat wearing either running shoes (RS) or weightlifting shoes (WLS). Young, healthy active adult males (*N*=20) were filmed in the sagittal plane while performing barbell back squats for each shoe condition at an intensity of 60% of one repetition maximum (1RM). While a number of kinematic parameters were similar between conditions, the shank maintained a more vertical position and the bar and hip were displaced less when wearing WLS, suggesting a more erect trunk posture. WLS may make small changes that allow for a safer, more effective squat performance.

KEYWORDS: footwear, power lifting, bar path.

INTRODUCTION: The traditional barbell back squat is both a competitive lift in the sport of powerlifting and an exercise commonly incorporated in the regimens of those who participate in resistance training or who are rehabilitating a lower extremity injury. The National Strength & Conditioning Association (NSCA) published a position statement and review article on the squat (Chandler & Stone, 1991). This position statement suggested that poor squatting technique could increase the risk of injuries, particularly to the knees and low back. To prevent such injuries, the NSCA recommends that the lifter maintain a normal lordotic posture and keep the torso as vertical as possible throughout the entire lift. The descent should begin with a slight forward bend at the hips while keeping the weight towards the heels, “sitting back” rather than shifting forward. More recent biomechanical studies have further measured joint kinetics and kinematics of the squat, from both performance and clinical perspectives, and these studies have reported joint range of motions (ROMs) (Kongsgaard et al., 2006), peak flexion angles of the lower extremity joints and segments (Fry et al., 2003; Salem et al., 2003; Flanagan & Salem, 2007), and displacement of the barbell (Donnelly, Berg, & Fiske, 2006) to evaluate the lifting technique.

Nearly all sports use some type of equipment or apparel aimed to help enhance performance and/or reduce injury risk, and this includes the sports of powerlifting and weightlifting. Weightlifting shoes (WLS) are designed with the intent to increase power production during Olympic-style lifts (i.e. the clean and jerk and the snatch) and squats; their main features are hard, incompressible soles that quickly redirect force upward from the floor, and raised heels to facilitate ankle mobility (Charniga; Kilgore & Rippetoe). Though no studies have investigated the effects of WLS or any other type of shoe, studies have shown that body-weight squats performed on a slight decline angle (similar to the effect created by WLS) have increased lower extremity muscle activity (Kongsgaard et. al., 2006; Richards et. al., 2008).

The purpose of the current study was to examine differences in squat kinematics when wearing running shoes (RS) and WLS. It was hypothesized that the WLS would alter the lower body joint ranges of motion (ROMs), allowing lifters to have a more erect trunk posture and perform a more efficient squat. Results from this study will add scientific understanding to the current anecdotal information of the biomechanical effect of WLS on barbell back squats.

METHODS: Healthy, college-aged males volunteered for this study (20±3 yrs.; 180±6 cm; 87±11 kg). All participants were relatively experienced in resistance training including the barbell back squat. After signing a consent form, each participant had an adequate amount of stretching and warm-up to replicate a regular training session. A 60-Hz Panasonic digital camera (Osaka, Japan) was placed approximately 1.3 m high and 5 m away on the left side to capture the two-dimensional barbell back squat kinematics in the sagittal plane. Reflective markers were placed on left side of the participant's fifth metatarsal joint (toe), lateral malleolus (ankle), lateral femoral epicondyle (knee), and greater trochanter (hip). An additional marker was placed on the end of the barbell. These five markers were used to create trunk, thigh, shank, and foot segments to calculate joint angles of the (a) hip, (b) knee, and (c) ankle, as seen in Figure 1. A segment from the hip to the end of the barbell was used to approximate the trunk segment since the end of the barbell is in the fixed position of the shoulder joint (McLaughlin, Dillman, & Lardner, 1977; Fry et al., 2003).

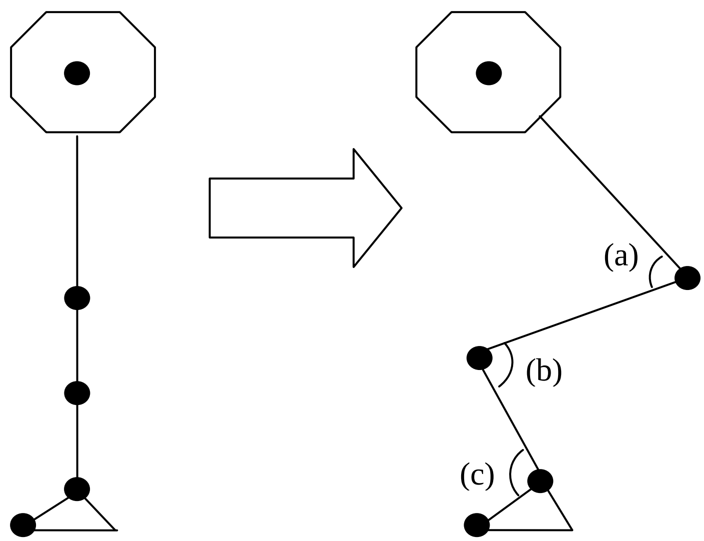

Figure 1: A diagram of the marker set up and joint angle measures.

As all participants were familiar with the barbell back squat, only a brief instruction was given to ensure the left foot was perpendicular to the camera position and the feet were pointed forward for proper tracking of the squat motion in the sagittal plane. If a participant felt uncomfortable with any aspect of the testing procedure, practice sets were offered. In order to achieve a comparable effort level from all participants, all trials were performed at 60% of their self-reported one repetition maximum (1RM). Each participant performed five repetitions using RS and five repetitions using WLS. All repetitions of a particular shoe type were completed together, but the order of shoe condition was randomized. For each squat repetition, participants began standing erect with the barbell on the upper back and descended until the thigh segment was roughly parallel to the floor, and then ascended back to the starting position.

The squat video was captured and the data were directly imported into Vicon Motus version 9.2 software (Centennial, CO). Two of the five repetitions were averaged and used for calculation purposes. The two repetitions chosen were the third and fourth repetitions in all participants. The kinematic variables measured were the posterior displacement of the hip and anterior displacement of the barbell from their initial positions, anterior displacement of the knee from the toe, ROM and peak flexion of the ankle, knee, and hip joints, and peak flexion of the trunk segment with respect to the vertical. The anterior bar displacement and posterior hip displacement were summed to create a variable termed "horizontal trunk displacement". Repeated-measures ANOVAs were conducted for horizontal trunk and anterior knee displacements, joint ROM, and joint peak flexion using Statistical Package for Social Sciences (SPSS) version 11.5 (Chicago, IL). To protect against Type I or II errors, α=0.05 and β=0.20.

RESULTS: Joint ROM, peak flexion angles, anterior bar and knee displacement values are shown in Table 1. The p value and statistical power are also reported. Significantly less horizontal trunk displacement was seen while lifters performed with WLS. This indicates the lifters maintained a more erect trunk posture during when squatting. There was also a statistically significant difference in ankle peak flexion, equating to a more vertical shank position, in the WLS condition. No other significant differences were seen between conditions.

Table 1. Comparison of squat kinematics between RS and WLS

Variable	RS	WLS	p-value	power
*Horizontal trunk displacement (mm)	231 ± 52	207 ± 43	0.04	0.55
Anterior knee displacement (mm)	90 ± 28	94 ± 30	0.44	0.12
Ankle ROM (deg)	23 ± 4	24 ± 5	0.17	0.27
Knee ROM (deg)	92 ± 13	93 ± 14	0.32	0.16
Hip ROM (deg)	96 ± 23	99 ± 15	0.44	0.12
*Ankle peak flexion (deg)	100 ± 8	103 ± 7	0.02	0.64
Knee peak flexion (deg)	81 ± 14	81 ± 15	0.83	0.06
Hip peak flexion (deg)	73 ± 15	73 ± 14	0.71	0.07
Trunk peak flexion (deg)	52 ± 9	52 ± 9	0.61	0.08

*Significant difference between shoe conditions, $p<0.05$.

DISCUSSION: The barbell back squat is an effective strength training exercise, when done in moderation with proper technique (Chandler & Stone, 1991). This study investigated whether WLS, specifically designed to improve squat technique, would significantly affect squat kinematics. Joint ROM and peak flexion angles were very similar between conditions, though a significant difference in ankle peak flexion indicated a slightly more vertical shank position, which is consistent with the teachings of proper squat technique (Chandler & Stone, 1991). With no difference in ankle ROM, it is likely that this is the direct effect of raised heel in WLS. The knees were able to move slightly over the toes in both conditions, which has been shown as an effective way to minimize hip and knee joint torque (Fry, Smith, & Schilling, 2003).

The biggest practical difference may have been in the horizontal trunk movement. From a physics perspective, the optimal bar path for the squat is a completely vertical line. In reality, there will always be some anterior bar displacement accompanied by some posterior hip displacement, creating a forward trunk lean. The goal, then, is to minimize these movements to reduce the amount of trunk lean. In a previous biomechanical analysis of elite lifters (McLaughlin, Dillman, & Lardner, 1977), higher skilled lifters had less trunk lean than their lower skilled counterparts. Accordingly, the NSCA's position paper also recommended minimal trunk lean to improve performance and reduce injury risk (Chandler & Stone, 1991). In this study, the combined amount of anterior bar movement and posterior hip movement was significantly less when wearing WLS. This seemingly corresponds to a more erect trunk posture, which coaches believe should reduce stress on the low back. Unfortunately, the population of this study did not exhibit any other kinematic differences when using WLS versus running shoes. However, during testing nearly all participants mentioned to the research staff that they felt the squats were much easier to perform when wearing WLS. The initial findings of this study do suggest that further research on the effects of WLS is warranted. Possible studies might include analyzing movement patterns throughout the various phases of the squat, incorporating higher loads, measuring kinetic variables such as peak joint torques and work, and tracking excursion of the center of pressure to consider stability levels. It is also suggested to integrate a larger, more diverse population of subjects to explore WLS effects' on squat mechanics with respect to age, gender, and training experience.

CONCLUSION: This is the first known study on the effects of footwear on squatting technique. Lifters demonstrated significantly more peak ankle flexion and significantly less combined anterior bar and posterior hip displacement when wearing weightlifting shoes (WLS) as compared to running shoes. This suggests that WLS allow for the more vertical shank position and erect posture during squatting that strength coaches recommend.

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BCH ANGLES OF YOUNG FEMALE WEIGHTLIFTERS DURING SNATCH MOVEMENT

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This study is aimed at using the BCH angle which represents the relative movement of lifter's body and barbell to compare the snatch movement of two young female weightlifters. The snatch movements of these two female weightlifters were filmed by a high speed camera (120Hz) and analyzed by Kwon 3D motion analysis software. The BCH angle, defined as the angle between the projection vector of the 7^{th} cervical spinous process to the barbell and the projection vector of the 7^{th} cervical spinous process to the hip joint in the sagittal plane, was calculated. The differences of BCH angles during the whole snatch movement have been found between successful and unsuccessful lifts for S1, and even between successful lifts of S1 and S2. In conclusion, the BCH angle seems to be a simple and good variable to evaluate the snatch techniques of the weightlifters.

KEYWORDS: barbell, successful lifts, film analysis

INTRODUCTION: The general kinematical characteristics of the barbell during the snatch for elite weightlifters have been determined in previous studies (Garhammer, 1985; Isaka et al.,1996; Baumann et al., 1998; Gourgoulis et al.,2000). Most studies acquired the kinematics of the barbell or the weightlifter's lower extremity joints under a single competitive or laboratory condition and eventually concluded that there was no parameter that was significantly different between successful and unsuccessful lifts (Stone et al., 1998). Furthermore, elite weightlifters seemed to have no one standard method to snatch the bar successfully (Chiu et al., 2007). The results of previous studies have shown that correcting some important bar path parameters by using visual and verbal feedback will not only improve the technique of the power clean or snatch for the lifters, but will also translate to increase power and force production in the athletes performing the lifts (Winchester et al., 2005; Winchester et al., 2009). In these studies, the verbal feedback was provided as to how to adjust body movement to obtain an optimal bar path as outlined by Stone et al. (1998). From the above studies, investigating the relative movement between the bar and the body of the lifter seemed to be more adequate to evaluate the technique of the snatch than alone analyzing the bar path or the kinematics of the body movement.
The BCH angle, a new single parameter which represents the relative movement of the lifter's body and the barbell, has been validated to characterize the snatch movements of an elite young female lifter (Chiu, et al., 2009). This study involved filming the snatch movement of two young female weightlifters and attempted to use the BCH angle to compare their snatch techniques.

METHODS: The snatch movements of two young female weightlifters were filmed on two different days. The physical characteristics and the lifted barbell mass of the two subjects are showed in Table1. Nine lifts of S1 (6 successful and 3 unsuccessful lifts) and eleven lifts of S2 (6 successful and 5 unsuccessful lifts) were analyzed and compared. This investigation was approved by the Human Experiment and Ethics Committee of the National Cheng Kung University Hospital. The subjects were informed of the experimental risks and signed an informed consent before participation.
A high-speed camera (Mega Speed MS1000, sampling rate: 120 Hz) was set on the left side of the lifters to film the snatch movement in the sagittal plane. A calibration rectangular plane (100cm long, and 140cm high) with 30 control points was used in this study. The two dimensional spatial coordinates of the selected points were calculated using a direct linear transformation procedure by Kwon 3D motion analysis software. The reconstruction errors were 0.21 and 0.25cm for the film analysis in the two days. The raw data was smoothed using

a 4th-order butterworth low-pass filter at a cut frequency of 6Hz. The barbell mass lifted was determined by the coach's instruction and the order was similar to that adopted in competitions.

Table 1. Physical Characteristics and the Lifted Barbell Mass of the Two Subjects.

Subject	Age (yrs)	Weight (kg)	Height (cm)	The barbell mass for the two different days (kg)
S1	19	54	157	1st: 80*, 80*, 83, 83*, 85, 85 2nd: 83*, 85*, 87*
S2	18	59	153	1st: 60*, 62*, 64*, 65, 65, 65 2nd: 62*, 63*, 64, 64, 64*

* Successful lifts

The six events defined in this study included: lifting the barbell off the floor (LO), clearing the barbell past the knee of the lifter (CK), extension of the lifter's hip joints to push the bar away from his body (PB), the barbell reaching its maximum forward position (MF), the barbell reaching its maximum vertical height (MH), and the lifter catching the bar overhead (CB). The BCH angle defined as the angle between the projection vector of the 7th cervical spinous process to the barbell and the projection vector of the 7th cervical spinous process to the hip joint in the sagittal plane was calculated. To take into account the fewer lifting numbers, independent t-test with SPSS statistical package was used, but only to compare the BCH angles of the two lifter's successful six lifts at the six events ($\alpha = 0.05$).

RESULTS: Table 2 shows the BCH angles of S1 and S2 in successful and unsuccessful lifts at the six events. The BCH angle of S2 was significantly greater than that of S1 only at the MF event. Although there is no statistical analysis, the mean BCH angles of S1 were smaller in unsuccessful lifts than those in successful lifts at the PB, MF and MH events.

Table 2. The BCH Angles (degree) of S1 and S2 in Successful and Unsuccessful Lifts at the Six Events.

Events	Successful lifts of S1（n=6）	Unsuccessful lifts of S1 (n=3)	Successful lifts of S2 (n=6)	Unsuccessful lifts of S2 (n=5)
LO	42.0 (4.5)	43.7 (2.0)	44.2 (3.1)	44.0 (2.5)
CK	34.7 (2.3)	35.5 (2.5)	36.0 (2.2)	37.3 (2.8)
PB	7.8 (5.0)	1.5 (1.5)	9.0 (6.14)	7.8 (5.8)
MF	27.1* (1.0)	23.8 (1.1)	59.0* (5.8)	60.6 (4.6)
MH	116.6 (4.5)	109.5 (3.7)	122.9 (6.7)	122.1 (8.3)
CB	198.5 (8.8)	208.6 (23.4)	194.9 (7.1)	192.9 (22.8)

* Significant difference between the successful lifts of S1 and S2 ($p < 0.05$).

After the CK event, the BCH angle decreased more slowly and the angular velocity rapidly decreased prior to the PB event in unsuccessful lifts for S1 (Figure1). However, the BCH angle of S2 decreased more rapidly after the CK event than S1 and there is also a marked decrease of angular velocity between the CK and the PB event (Figure 2). Approaching the

MH event, there is a maximum BCH angular velocity. It is obvious that the peak BCH angular velocity was greater in unsuccessful lifts for S1 (Figure1).

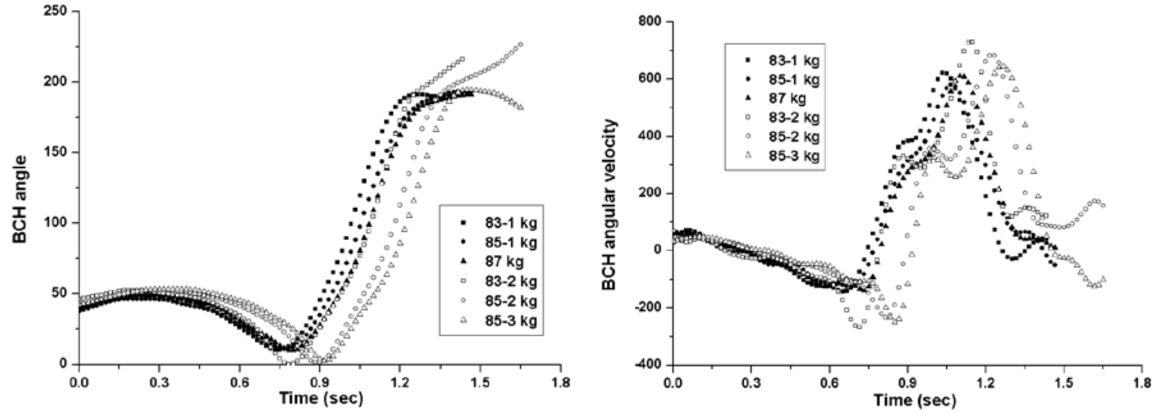

Figure 1. The BCH angle (degree) and angular velocity (deg/s) during successful lifts (■,●,▲) and unsuccessful lifts (□,○,△) for S1.

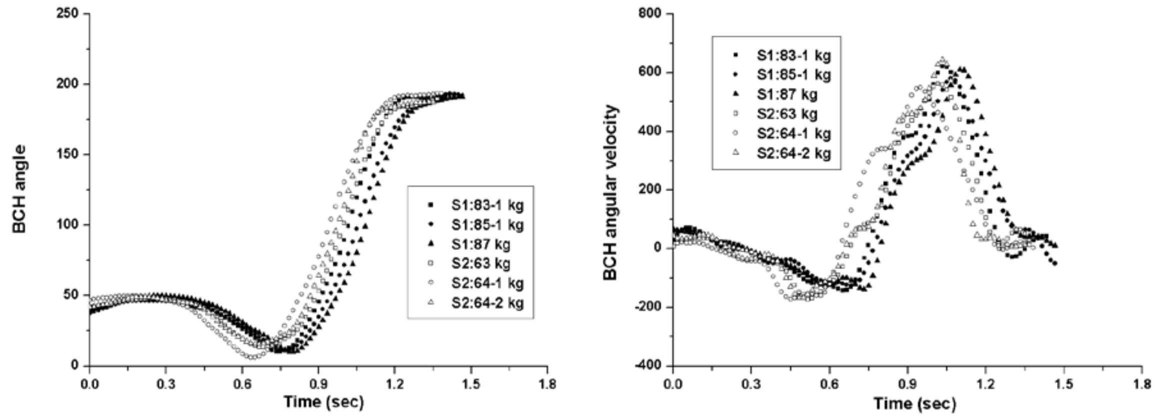

Figure 2. The BCH angle (degree) and angular velocity (deg/s) during successful lifts for S1 and S2.

DISCUSSION: This present study attempted to use the BCH angle to evaluate the snatch technique. In previous studies (Stone et al., 1998), there were no parameters significantly different between the successful and unsuccessful lifts. However, differences between the BCH angles of S1 during the whole snatch movement have been found between successful and unsuccessful lifts in this study (Figure 1). The previous study showed that the female lifter performed with the same BCH angles from the LO to the CK event as lifting a barbell of 80～82 kg (Chiu and Liang, 2009). In this present study, this female lifter also had a stable performance from the LO to the CK event. However, after the CK event, her BCH angles decreased slowly in unsuccessful lifts. Even though the rapidly following reduced BCH angles were performed to compensate for the prior slower movement, the greater BCH angular velocity at the MH event caused the bar to drop backward.

In this study, the two female subjects had different snatch techniques. Subject 1 pulled the bar with a jump backwards resulting in a catch position farther behind the initial position of the barbell than subject 2 who caught the bar in the initial position or slightly forward. However, the pattern of the BCH angle and the angular velocity curves during the whole snatch movement for S2 were similar to S1. The difference between the snatch techniques of the two lifters perhaps is that subject 2 had rapid decreased BCH angles after the CK event. Eventually, she pushed the bar earlier or perhaps at a lower position to her thighs and that caused a greater BCH angle at the MF event (Table2). The mean greater BCH angle (59.0±5.8 degree) at the instant the barbell reached maximum forward position means that the lifter's upper body is farther away from the barbell. This will increase the difficulty in catching the bar in time as the bar drops down.

CONCLUSION: In this study, the differences of BCH angles during the whole snatch movement have been found between successful and unsuccessful lifts for S1, and even between the successful lifts of S1 and S2. The BCH angle seems to be a simple and good variable to evaluate the snatch techniques of the weightlifters. Based on the results, subject 1 should avoid the slowing of the decreased BCH angle as she is pulling a heavier barbell, especially from the CK to the PB event. Conversely, subject 2 should slow her pulling bar movement and push the bar from the upper position of her thighs.

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Oral Session 03

Modeling

A COMPUTATIONAL MODEL TO INVESTIGATE SHOE AND SHOE-SURFACE INTERFACE EFFECTS ON ANKLE LIGAMENT STRAINS DURING A SIMULATED SIDESTEP CUTTING TASK

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Ankle sprains account for 10% to 15% of reported sports injuries. High ankle sprains are currently thought due to torsional loads and potentially debilitating to the athlete. In the current study a computational model was developed to investigate the human response in shoes on different athletic playing surfaces during a simulated sidestep cutting task. Ankle ligament strains were obtained from the model to help predict ankle injury. The model may provide a computational basis for studying shoes and shoe-surface interfaces that can be used to help optimize player performance and minimize injury risk.

KEYWORDS: ligament, ankle injury, foot constraint, shoe-surface interface, model.

INTRODUCTION: Ankle sprains are one of the most frequent injuries in sports and often account for 10% to 15% of reported injuries (Villwock et al. 2009a). In the National Football League high ankle sprains, also known as syndesmotic sprains, constitute approximately 20% of ankle injuries (Boytim et al. 1991). These injuries receive considerable attention due to their relatively long recovery time. While high ankle torque is implicated as a risk factor, a cadaver study by Villwock et al. (2009a) suggests foot rotation may be a better predictor of injury. In addition, Villwock et al. (2009b) investigated the effects of various cleated shoe designs and playing surfaces on torsional responses using a surrogate ankle. Their study shows that synthetic surfaces yield higher torques and rotational stiffnesses than natural grass surfaces. The study also suggests that a shoe with a pliant upper, allowing more subtalar motion, may provide less foot constraint than a rigid upper, and consequently help protect the foot from rotational injuries.
The objective of current study was to develop a computational model, guided by the above cadaver and playing surface studies, to investigate the effects of foot constraint and shoe-surface interface on ankle ligament strains. The model may have utility in providing a computational basis for shoe and playing surface designs that will optimize player performance and minimize injury risk.

METHOD: Joint anatomical features were taken from one computed tomography (CT) set in the cadaver studies (Villwock et al. 2009a). Detailed features of the ankle were obtained by importing Digital Imaging and Communications in Medicine (DICOM) files of individual CT scans into Materialise's Interactive Medical Imaging Control System (MIMICS) (Materialise, Ann Arbor, MI). This yielded a three-dimensional surface model of the bones as Stereolithography (STL) files. To reduce the size of the surface files and subsequent model, the STL files were remeshed in MIMICS to smooth the surfaces of each bone. Exported files were then imported into the 3-D solid modelling software SolidWorks (TriMech Solutions, LLC, Columbia, MD) as Mesh Files (.stl) (Liacouras et al. 2007). SolidWorks, along with its ScanTo3D package, was used to further construct each bone and simplify the bone surfaces. A rigid plane (200 x 100 x 10 mm) was created in SolidWorks to represent the playing surface. The SolidWorks Motion package was then used to assemble the bones and surface, obtain proper positioning, add necessary components, and run simulations.
The ligaments were represented as linear springs (Fig 1), and their stiffness values, origins and insertions were based on the literature (Liacouras et al. 2007; Netter et al. 2003). The foot model was composed of 14 bones. A compressive pre-load of 1600 N, approximately 2X BW, was distributed between the tibia and fibula. Two kinds of foot constraint were

simulated, representing rigid and pliant shoes. The rigid shoe model had 3 contacts and 10 ligaments. The tibia was fixed in space, and the fibula and remaining bones (acting as a rigid body) were free to move (Fig 1 left). The pliant shoe model had 6 contacts and 12 ligaments. The tibia was fixed in space, and the fibula, calcaneus and the remaining bones (acting as a rigid body) were free to move, allowing subtalar motion (Fig 1 right).

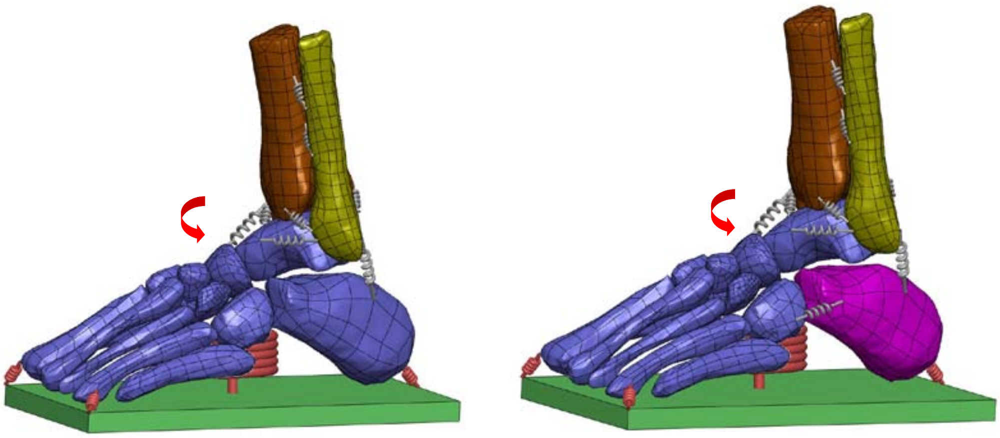

Figure 1. Foot and surface models. Left: rigid shoe; Right: pliant shoe.

The shoe-surface interface was simulated with a torsional spring at the surface center and 3 lateral surface springs, as shown in Figure 1 (red springs). Two kinds of surfaces, synthetic turf and natural grass, were simulated with stiffnesses of 3.6 Nm/deg and 2.2 Nm/deg, respectively, based on the literature (Villwock et al. 2009b). The stiffness of the 3 lateral surface springs was 1 N/mm. A rotary motor was applied to the surface aligned at the ankle center between the malleoli. The external rotation angle was set to 30° to simulate a common sidestep cutting task (Dowling et al. 2010).

RESULTS: Critical ligament strains were measured in the computational models for various test conditions. For clarity, only strains of the ligaments injured in the previous cadaver experiments (Villwock et al. 2009a) were plotted in Figure 2 and Figure 3; namely, the posterior talofibular ligament (PTaFL), the deltoid ligament (representing the anterior tibiotalar ligament), and the anterior tibiofibular ligament (ATiFL).

The maximum strains in each ligament were greater on the synthetic turf than on natural grass, and the order, from the highest to the lowest, was rigid shoe on turf, pliant shoe on turf, rigid shoe on grass, and pliant shoe on grass (Fig 2). This indicated that shoe-surface interface influenced ankle ligament strain and consequently the potential for ankle injury. For the rigid shoe condition, the maximum strain occurred in the PTaFL, while the deltoid ligament experienced the highest strain in the pliant shoe, suggesting that the location of ankle injury might depend on foot constraint. On the same surface, higher ligament strains were developed for the rigid than pliant shoe, indicating that less rotation might be required to fail ligaments with a rigid upper design. Finally, two large changes in ligament strain were noted in the PTaFL, namely, from Rigid/Turf to Pliant/Turf and from Rigid/Grass to Pliant/Grass. Apparently, these were due to foot constraint, and therefore, the effect of foot constraint on ligament strain seemed to be greater for the PTaFL than the other two ligaments. Comparably, significant changes due to surface were noted in both the PTaFL and the deltoid ligament, such as from Rigid/Turf to Rigid/Grass and from Pliant/Turf to Pliant/Grass. This finding indicated that shoe-surface interface affected strains in the PTaFL and deltoid ligament in a similar fashion.

The strain-rotation behaviours of each ligament were plotted in Figure 3. While linear springs were used to represent ligaments in these models, the strain-rotation behaviour was non-linear due to the complex joint geometry.

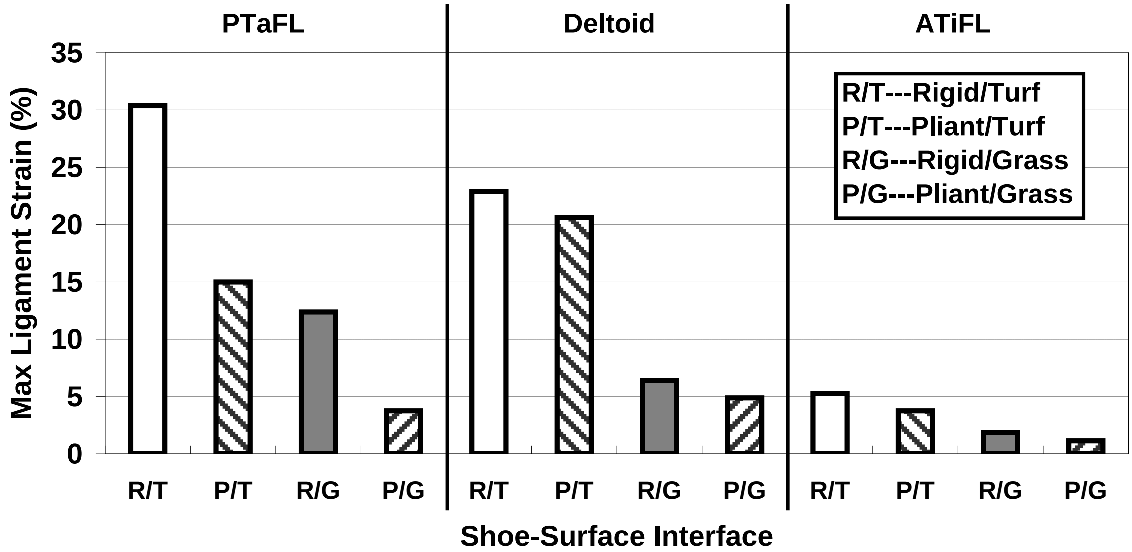

Figure 2. Strains in various ligaments at 30° of external surface rotation for different shoes and shoe-surface interface conditions.

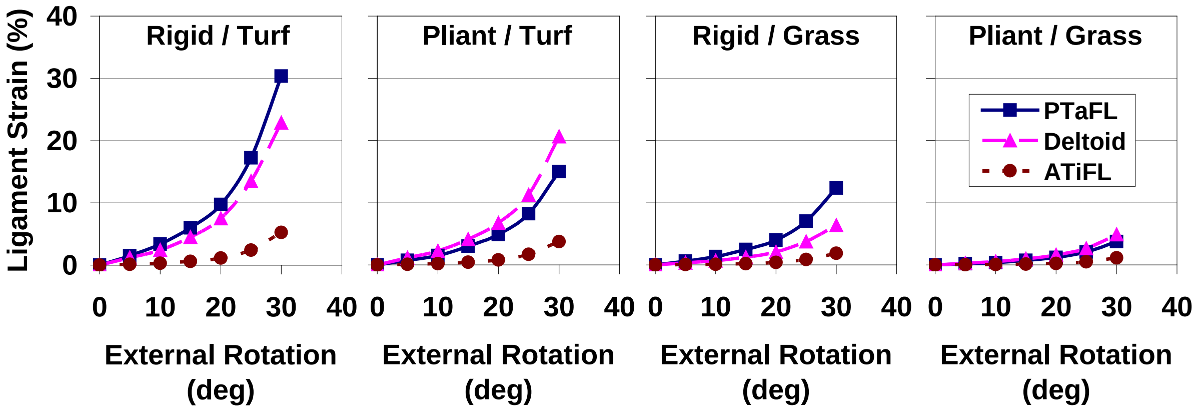

Figure 3. Strains in various ligaments during external rotation of the surface for different shoes and shoe-surface interface conditions.

DISCUSSION: The results from the current models were consistent with ligament injuries in the cadaver study (Villwock et al. 2009a), suggesting that ankle injury depends on foot constraint. Models of the shoe and shoe-surface interface conditions in the current study indicated that larger strains were developed in ankle ligaments on synthetic turf than on natural grass for a simulated sidestep cutting task.

Subtalar motion of the foot was allowed in the pliant shoe model by freeing up the calcaneus (Fig 1 right). Consequently, ankle ligaments in the pliant shoe experienced less strain than in the rigid shoe on the same surface. This agreed with the Villwock et al. (2009b) study suggesting that a pliant shoe upper may help reduce the risk of ankle injury. For performance purposes, however, football and soccer players tend to wear tight fitting shoes with relatively rigid uppers to provide more foot constraint. An in vivo study by Dowling et al. (2010) showed that during the sidestep cutting task, subjects were able to obtain the desired cut of approximately 30° on the high friction surface but only 24° on the low friction surface. Their study, together with Villwock et al. (2009b), suggests that synthetic surfaces with high friction may benefit player performance. Yet, according to the current study, the synthetic turf may compromise ankle mechanics and increase the risk of injury.

A recent study by Drakos et al. (2010) investigated the effect of different shoe-surface combinations on ACL strain and suggests that a cleat-grass interface may result in fewer noncontact ACL injuries than the turf shoe-turf interface. Similarly, Dowling et al. (2010) suggests high friction synthetic turfs may be associated with an increased incidence of ACL

injury. While the current study documented only ankle ligament strains, the results support the notion that synthetic surfaces may be a potential risk factor for rotational injuries of the lower extremity, especially when using a rigid shoe design.
Some limitations of the model should be noted. A study by Beumer et al. (2003) determined the strength and stiffness of the tibiofibular and tibiotalar ligaments of the ankle and showed no differences between these ligaments having an average strength and stiffness of 550 N and 98 N/mm, respectively. Ligament elongation at failure was estimated to be approximately 5.6 mm. Based on an estimated length of these ligaments from the current study, this level of elongation may suggest failure strains on the order of 25% to 31%. While this level of strain was predicted by the rigid shoe and synthetic turf condition for the simulated sidestep cutting task, our model did not include any nonlinearity in response of the ligaments. We would suggest future simulations incorporate this well documented nonlinear response of ankle ligaments for failure prediction studies. While cleat pattern at the shoe-surface interface is an important factor in shoe design, its effect on failure analyses is yet unknown and should be studied in the future.

CONCLUSION: Computational models of shoe and shoe-surface interface conditions were developed to measure ankle ligament strains during a simulated sidestep cutting task. The results showed that the maximum ankle ligament strains were generated with a rigid shoe model on synthetic turf. The ability of the current models to incorporate various shoe-surface interface characteristics and show differences in predicted ankle ligament strains was encouraging. While more failure data are needed on ankle ligaments, additional experiments using human cadavers and in vivo tests with human subjects will also be needed to validate computation models. Ultimately, such models may provide a basis for optimizing shoe designs and shoe-surface interface characteristics to enhance player performance and minimize injury risk.

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A NEW METHOD FOR UNCONSTRAINED MEASUREMENT OF KNEE JOINT ANGLE AND TIMING IN ALPINE SKIING : COMPARISON OF CROSSOVER AND CROSSUNDER TURNS

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Timing is an important part in the analysis of alpine skiing technique. Key events extracted from dynamics and kinematics of ski were previously proposed, but timing of body segments kinematics is also required to understand the biomechanics and to assess the performance. In this study, we proposed a new method based on inertial sensors to measure timing parameters based on body segments kinematics. To show the efficacy of the method, angle and timing of knee joints during crossover and crossunder techniques were compared. Significant differences were obtained for the timing, but not for the amplitude of the knee angles. The proposed system reported a good repeatability, did not encumber the athletes and allowed the measurement of body movement during the whole run. So it could easily be used to assess performance in training conditions.

KEYWORDS: Alpine skiing, inertial sensors, timing, knee angle, training conditions

INTRODUCTION: Measuring the timing of key events of alpine skiing technique (e.g., beginning or end of steering phase) is essential to understand the biomechanics and performance of this winter sport. For example, when comparing carving and traditional techniques, Müller & Schwameder (2003) noticed important differences in the timing periods of the initiation and steering phases. Currently, the key events timed in alpine skiing are based on the dynamics (Vodickova & Vaverka, 2009) or kinematics (Wägli et al., 2009) of the skis. Although crucial, these events are not sufficient for a complete characterization. In fact, it is also necessary to measure the key events of body segments kinematics.
Video camera is the common method for timing evaluation. According to Chardonnens et al. (2009), it is a precise measurement system (about two images or 8 ms with a standard 2D 25Hz camera). However, this method requires many manual operations and is therefore time consuming. Three-dimensional (3D) optical-based systems have also been used to quantify skier timing. Nevertheless, due to their technical complexity (e.g., small capture volume, requirement of professional staff for the setting-up and post-processing), these systems are limited to research studies. Consequently, there is a need for an easy-to-use timing system which can be directly used by trainers under in-field conditions. Wearable systems composed of inertial sensors were recently proposed as an alternative, suitable for routine evaluation. For alpine skiing, methods were proposed to measure body segment kinematics (Supej, 2009) and ski kinematics (Wägli et al., 2009). In addition, a system was presented to time temporal events during ski jumping (Aminian et al., 2009). Although promising, these methods require further improvement and validation.
The aim of this study was to propose an inertial-based method to automatically measure the relative knees joint angle and time particular events without limitation regarding the duration of the captures or their locations. In order to show the efficacy of the proposed method for alpine skiing, knee joint angular and timing parameters were defined and compared between crossover and crossunder techniques.

METHOD: A wearable system composed of four wireless inertial modules (Physilog®, BioAGM, CH), fixed laterally at the middle length of the thighs and behind ski boots, was designed to measure knee joints kinematics. Each module, weighting less than 100g, was composed of a 3D gyroscope (±1200 °/s), a 3D accelerometer (±10g) and an embedded datalogger recording the signals at 500 Hz. Before each run, a functional calibration procedure was performed to align the modules with the segments as proposed in Favre et al.

(2009). In order to evaluate the repeatability of the method, the inertial-based system was removed and replaced between each run (test-retest).
Six alpine skiers (27±4 years old, 180±5cm, 78±10kg), among which three were professional instructors and three were experienced skiers, took part in this study. They performed twice crossover (CO) and twice crossunder (CU) techniques on a regular slope (0.4km long, 100m large, 40% steep) with their own carving skis (13.5±0.8m radius; 166±8cm length). In addition, a standard video camera recording at 25Hz and synchronized with the proposed system was used to record the runs. Snow and weather conditions were similar for each skier.
The major difference between these two skiing techniques is that in CO technique, to edge from one side to the other during the initiation phase, the trunk swings over the lower limbs. On the contrary, in CU technique, the trunk remains stable and the lower limbs cause the change of edges. Based the knee joints function for these techniques, four events were defined for each turn as described in Table 1 and illustrated in Fig.1. These timing events were determined automatically for each turn using a robust detection method based on the differential sagittal angular velocity between thighs and shanks. The flexion-extension relative angles of knees modeled as a hinge joint were also calculated based on the angular velocities and accelerations measured by the four modules and by considering some joint and turn constraints (Bortz, 1971).

Table 1. Timing events definition from knee kinematics

BE	Beginning of the extension for the outer knee
BF	Beginning of the flexion for the inner knee
EE	End of the extension for the outer knee
EF	End of the flexion for the inner knee

In order to compare the two techniques, the range of knee flexion-extension, the absolute duration of the turn, as well as four intra-turn durations ($\Delta B = BF - BE$, $\Delta E = EF - EE$, $\Delta Ex = BE - BE$ and $\Delta Fl = EF - BF$) were calculated for each turn. The duration of the turn was defined as the time between two consecutive *BF*. The intra-turn timing periods were expressed both, as absolute values (in ms) and as percentage of the turn duration. The two first and last turns of the run were considered as transition phases and were therefore ignored in this analysis. Wilcoxon rank sum test was used to determine the level of significance of the differences observed between the two techniques. This test was done for the range of motion, the turn duration, as well as for the durations (expressed in ms and as percentage of the turn duration) considering all skiers and all turns of the two runs. The repeatability was assessed by Wilcoxon rank sum tests. In this case, the angular and timing parameters were compared between the first and second run. These tests were done separately for the CO and CU techniques, in each case considering all turns of the six skiers. An alpha value of 0.05 was considered to assess statistical significance.

RESULTS: An average of 11±4 turns per run was collected for each athlete. In total, the relative knee angles and the respective timing periods were calculated for more than 100 turns per technique. Typical results are presented for the CO and CU techniques in Fig.1.

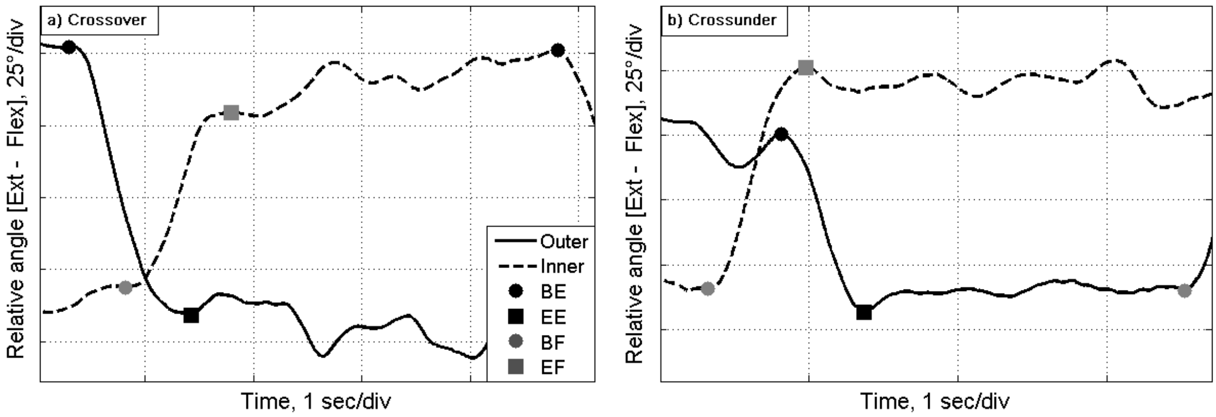

Figure 1. Typical relative knee flexion-extension angles and their temporal features a) crossover CO and b) crossunder CU techniques.

When comparing CO and CU techniques considering all turns, non significant differences were obtained for the range of motion and for the absolute timing periods. However, the timing periods expressed in percentage of the turn duration showed significant differences. Indeed, in this case, we found that *ΔB* and *Δ*E were significantly lower for CU than for CO. On the contrary, a significant increase was obtained between *ΔEx* and *ΔFl* when comparing CU to CO. Finally, although CU generally reported a longer turn duration than *CO*, this difference was not significant (p=0.07). The medians of the timing periods expressed as percentage of the turn duration are presented in Table 2.

Table 2. Medians in % of turn cycle for CO and CU

	ΔB	*ΔE*	*ΔEx*	*ΔFl*
CO	28	44	20	16
CU	-30	-34	50	70

The test-retest differences expressing the repeatability of the method were also not significant for the knee range of motion and all timing periods in ms and in percentage with the CU and CO techniques, except for *ΔB* in ms (p=0.03) with CU.

DISCUSSION: The wearable system presented in this study allowed the measurement of the relative knees angle and a robust detection of timing events based on 3D acceleration and 3D angular velocity measured at the thighs and boots level. It is worth mentioning that the angular and timing parameters were automatically calculated.
The proposed method was reliable on both CO and CU runs, since no significant changes were observed when the system was removed and replaced. Moreover, by showing significant differences between the timing periods of both techniques, this method showed a good efficiency to study skiing biomechanics.
It is interesting to note that significant differences of the timing periods between both techniques were only observed when the timing were expressed as percentage of the turn duration. This can be explained by the fact that the duration of the turn is different for each turn and by the fact that some temporal features are proportional to the turn duration. Therefore, by calculating their values as percentage of the turn duration, this proportionality was highlighted. In other words, this suggests that within each technique, the considered timing periods (*ΔB, ΔE, ΔEx, ΔFl*) are proportional to the turn duration and do not have an absolute constant duration.
Regarding the timing analysis of CU, we reported that the initiation of knee movement corresponded to the beginning of the outer leg knee flexion (*BF)*. The beginning of the

extension movement (*BE)* appeared after *BF*, as illustrated in Table 2*.* Thus, no vertical movement of trunk was created. On the other hand for CO a positive ΔB was obtained, meaning that the movement was initiated by inner leg. The knee extension of the inner leg appeared before the flexion of outer leg, which induces a vertical displacement of trunk. A positive period ΔE for CO and negative for CU can be explained in the same way. Fig. 1 clearly illustrates this phenomenon.
Longer extension ΔEx and flexion ΔFl durations were reported for CU compared to CO technique (Table 1). This could be explained by the fact that swinging the legs under the trunk could require a larger percentage of the turn cycle than swinging the trunk over the legs, since lower limbs induce the entire movement. It is also important to note that CU is less commonly practiced than CO technique.

CONCLUSION: In this study, we proposed a new automatic method for the measurement of knee joint relative angles and timing periods based on inertial sensors. The proposed system did not encumber the subject and could easily be used during daily training. It allowed identifying relevant intra-turn parameters, as well as parameters relative to the whole run. The system showed a high sensitivity regarding the timing periods in term of percentage, as well as a good repeatability.
In addition, this study stressed the fact that relative angular amplitudes are not sufficient to characterize crossover and crossunder techniques. Indeed, significant differences between these techniques were only observed for relative timing durations. Although inertial sensors were sufficient to analyze the timing of knee, force and trajectory measurement could be added to complete the analysis. As skiing involves the movements of all body segments, other joints (e.g., hips) could also be considered in order to have a more general picture of the joint kinematics timing. Considering that the proposed method could easily be adapted for other joints, it was concluded that it has a high potential for daily monitoring of alpine skiing.

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CREATION OF THEORETICAL DATA SETS TO EXAMINE MOVEMENT VARIABILITY USING MODELLING

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KEYWORDS: golf, outcome measure prediction, pseudo-random data, simulation

INTRODUCTION: Recently, a large amount of research has been focused on the effect of movement variability on human performance in sport. It is now generally accepted that specific amounts of variability are essential to attain a high level of performance (Davids *et al.*, 2003). When studying the effect of movement variability on outcome performance, the usual method involves collecting numerous data sets from an individual and, assuming that these data sets will all be different (i.e. contain variability), attempt to connect the amount of variability to the change in outcome or performance measure using a number of statistical techniques. The aim of this study is to remove the requirement to collect a large amount of data which, by chance, may contain the level of variability required and shorten the data collection phase significantly by using the proposed process to create theoretical data sets containing alterable variability content while still exhibiting major characteristics of the actual data. When these theoretical data sets are used in conjunction with a full-body 3D computer model operating inverse and forward dynamics simulations a change in outcome or performance measure can be predicted. The advantages this process offers over traditional techniques is the ability to directly control and quantify the amount of variability introduced into the test data and a significant reduction in data collection time.

METHOD: Initially, a full-body, 42 degrees of freedom 3D computer model was created and validated using single-subject analysis. One elite female golfer (handicap 0) performed 12 shots with her own driver club. A 6-camera MotionAnalysis infrared camera system operating at 400 Hz recorded the kinematic data of the 27 markers located on the subject and this data were used to drive the computer model in ADAMS/LifeMOD software; model construction methods closely followed that of Nesbit (2005) and kinematic validation replicated Kenny *et al.* (2008). The results illustrate a high level of correlation (r^2=0.90) between the kinematic data collected in experimentation and the predicted trajectory of the validation markers of the model. The long-term focus of this work is on the effect of variability at one joint and the resultant change in both outcome measure and kinematics of other joints. However, the first stage is to create the theoretical data sets. To ensure the amount of variability within the theoretical data sets were controlled and realistic, the original data were analysed and used as the base data set. All 12 trials were used - the right knee angle data were all normalised to 101 points, a mean ensemble curve was created from the base data sets and the average standard deviation (sd_{avg}) occurring over the whole trial was calculated. The average standard deviation was used to signify the average amount of naturally occurring variability in the standardised trial data, i.e. variability not caused by an external factor such as fatigue. Variability was added to the mean ensemble curve at 20 different levels, the maximum variability curve was created by adding a random number between $\pm sd_{avg}$ to each data point; as the random number had containment limits it is considered pseudo-random only. Other data sets were created by reducing the pseudo-random number magnitude in 5% decrements to a minimum of 5% sd_{avg}. As a result 20 data sets were created each with differing variability content; set one ±100% sd_{avg}, set two ±95% sd_{avg} etc. As the random number is based on white noise (having a distribution with mean and median of zero), the data occurring at this intermediate stage was not representative of the main characteristics of the base data due to relatively large rates of change between consecutive data points. To remove these inconsistencies all 20 data sets were filtered using a 4th order reverse pass Butterworth filter with a cut off at 12Hz (a cut-off which has been reported to be useful for golf related data – Mitchell *et al*., 2003). The filter was not optimised for each data set as it was not the intention to remove the noise, only reduce the issue related to rate of change. As a result, 20 data sets were created each with a different amount of variability imposed on the base data. This variability was based on the characteristics of the original 12 data sets and as such are proposed to be representative and realistic data sets.

Due to the nature of the white noise based pseudo-random data it is essential to examine if the theoretical data sets follow the proposed pattern, e.g. does the data set based on ±65% sd_{avg} exhibit more variability than that based on ±45% sd_{avg}. To do this a Bland-Altman analysis (B&A) was completed; B&A is used to compare two measurements of the same variable. As the data presented here is time normalised each data point on the theoretical data set has a corresponding data point on the mean ensemble curve and is therefore considered a valid method of comparison. The 95% limits of agreement (LOA) from the B&A analysis will be used to assess the amount of variability contained within each theoretical data set and the bias will be used to assess if the gross pattern of the mean ensemble curve has been altered.

RESULTS AND DISCUSSION: The B&A analysis indicates that the LOA reduces as less variability is added to the data; from 1.44 at ±100% sd_{avg} to 0.267 at ±5% sd_{avg} (see Table 1). Further analysis reports an r^2 of 0.9264 when correlating the LOA values and the magnitude of the random number. The bias remains close of zero on each curve, indicating that the variability is equally distributed above and below the ensemble curve.

Table 1: Bland & Altman Analysis Results for Altered Levels of % of sd_{avg}

% of sd_{avg}	Bias (^{0})	LOA (^{0})	% of sd_{avg}	Bias (^{0})	LOA (^{0})	% of sd_{avg}	Bias (^{0})	LOA (^{0})
5	-0.016	0.267	40	-0.006	0.533	75	-0.093	0.863
10	-0.006	0.277	45	-0.016	0.620	80	0.048	0.922
15	0.010	0.249	50	0.019	0.737	85	0.108	1.161
20	-0.021	0.380	55	0.092	0.811	90	0.155	1.147
25	-0.025	0.360	60	0.048	0.767	95	-0.202	1.230
30	0.004	0.481	65	-0.014	0.739	100	0.006	1.444
35	-0.051	0.475	70	0.068	0.664			

CONCLUSION: The method outlined here, utilising a mean ensemble curve in conjunction with the addition of pseudo-random data and Butterworth filtering enables the practitioner to create valid and representative theoretical data sets which do not remove the main characteristics of the original data sets; as illustrated by the Bland & Altman analysis. The combination of these theoretical data sets, where the amount of variability can be controlled, with a full body 3D computer model of the golf swing leads to the ability to assess the impact of variability on both performance and outcome measures within human movement without having to acquire large amounts of data. The combination of these techniques expedites the reporting process within a sports setting, and allows a dramatic reduction in subject involvement during initial data acquisition compared with more traditional methodologies. Future work will concentrate on the effect the variability of joint angles has on the outcome measures, e.g. ball speed, ball carry, spin rates and performance measures within golf, e.g. weight shift patterns, x-factor stretch and swing plane. The research will further assist biomechanists in assessing the impact of levels of variability on human movement.

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METHOD TO VISUALIZE AND ANALYZE SIMILARITIES OF MOVEMENTS – USING THE EXAMPLE OF KARATE KICKS

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Most sports movements are very complex. In order to estimate similarities between the movements many biomechanical parameters are necessary. This impairs the clearness of the classification of similar movements. Another variant is the use of nonlinear procedures which allow the consideration of the movement on the whole. But a connection to the single biomechanical parameters is not given. The aim of this paper is to introduce a procedure to visualize the movement coordination to get a visual impression of the whole movement, and in addition further analyses by means of statistical tools to confirm similarities and variabilities between the movements were presented using the example of Mae-Geri.

KEYWORDS: coordination, karate, visualization, movement similarity

INTRODUCTION: Because sports movements are very complex, biomechanical analyses contain many parameters including their characteristic curves. For this reason it is frequently problematic to identify similarities and differences between the movements. Yet, the sports practice requires the identification of movement modifications. Therefore, various holistic approaches and methods have been used (Haken 1996; Yamada, 1995; Newell et al., 2007; Perl 2004; Schmidt et al., 2009; Witte et al., 2003; Witte et al., 2009). The problem with the recent methods is that they demand a decision between the analysis of the whole movement and movement details in the form of biomechanical curves or parameters. Based on this, it is the aim of this study to present a method which meets the following requirements: subjective impression of the total movement coordination with a visual pattern, quantitative analysis of biomechanical parameters and curves and application of statistical methods or tools to find similarities and differences between the movements.

METHOD: The generation of the visual movement patterns includes the following steps: selection and normalization of biomechanical parameters relevant to movement (e.g. body angles, angular velocities, and forces) temporal normalization, construction of a matrix containing movement parameters in discrete time-lags, and visualization by means of contour plots in colour or gray scales. After this, statistical methods or tools can be applied to the biomechanical time series. The researched Mae-Geri is a front kick which belongs to the karate sport (Figure 1). Five karatekas (age between 13 and 18 years) of high national level participated in this study: two male subjects: Chr and Joh and three female subjects: Lui, Mar and Nad. Each technique was performed ten times. Chudan (jap. Solar plexus) was defined as the target area. The movement analysis was accomplished with a VICON system (12 cameras MX 13, 250 Hz, Nexus V 1.01). For the movement patterns, the time courses of body angles were exported. Table 1 shows an overview of the movement specific body angles. The selection resulted from our own empirical findings and practical experiences of long-time karatekas. The angle normalization can be accomplished with the following determinants: angle maximum is 0 and angle minimum is 1. For the absolute time scale of each movement the following temporal standardization was computed: 0.0, 0.1, 0.2,…, 1.0. In this paper the results of the subjects Chr, Lui and Nad are presented. Afterwards, a matrix containing movement parameters in discrete time-lags was constructed and visualized by contour plots in gray scales. To research similarities between the movements Euclidean distances were calculated. To estimate similarities between the single trials and the single angles, coefficients of variation and correlation (Pearson) and the Euclidean Distance were used. Euclidean Distance for the single angles between the executions for each subject were calculated and similarity levels were defined (Table 2).

Figure 1. Technique of the Mae-Geri Chudan

Table 1. Body angles of Mae-Geri using for the movement pattern plots and statistical analyses recording to the Plug-in-Gait model by VICON

Short cut	angle
W1	HipX: Hip angle (Flexion)
W2	Knee: Knee angle (Flexion)
W3	Ankle: Ankle angle (Dorsiflexion)
W4	Spine: Spine angle (Flexion/Dorsiflexion)
W5	PelX: Dorsiflexion of pelvic
W6	PelZ: Internal Rotation of pelvic

Table 2. Definition of similarity levels on the basis of the Euclidean Distance

Similarity level	Range of the Euclidean Distance
I	0,0 – 0,2
II	>0,2 – 0,4
III	>0,4 – 0,6
IV	>0,6 – 0,8
V	>0,8 – 1,0
VI	>1,0

RESULTS: Figure 2 shows two examples for the three subjects for visual movement patterns as contour plots. At first sight and with respect to the temporal successions of gray scales, substantial similarities are visible for the subjects Chr and Lui. For example, the subjects Chr and Lui exhibit analogical characteristics for the first angle W1 (Hip. Greater differences within an athlete are observable for the angle PelX (W5) for the subject Nad. Generally, in comparison to the other subjects, more variable movement patterns can be found for the athlete Nad. To quantify the body angles for each subject, coefficients of variation were calculated as shown in Figure 3. The procedure was that at each normalized time point (in sum 11 time points) per angle, the coefficient of variation was computed. The time-averaged coefficient of variation for each angle and each subject is presented in Figure 2. It can be detected that the angles for each subject have a different behaviour of variability. Thus, for the athlete Chr the angle of the ankle (W3) is the most stable one. In contrast, Lui and Nad displayed the most stable knee angle (W2). This implies that the ankle and knee angle can be replicated the most. As distinguished from these the PelX (W5) for Chr and Nad is the angle with the highest variability. When comparing all coefficients of variation between the subjects, it can be concluded that the kicks of Lui show the least variability and the most variable movements can be found for Nad. The next outstanding problem was to detect which temporal angle curves show the greatest similarities. Therefore the coefficients of correlation (Pearson) between all five subjects for each angle were determined and from this the mean values were calculated. It could be found that the lowest mean correlations (the greatest variations) exist for the angles PelX W5 (r=0.16), Ankle W3 (r=0.75) and Spine W4 (r=0.73). From this it can be assumed that for the Mae-Geri the hip angle, the knee angle and the rotation of pelvis (here only a small amplitude is realized) are the most important angles. As a further method to identify similarities, the determination of the Euclidean Distance was used. From this the most stable angle for a subject can be concluded. In Figure 4 the percentage frequencies of the appearance of the similarity levels for each angle are represented. So it can be established that for Chr and Lui the similarity level I for W1 occurs most frequently. W1 (Hip) is characterized by relatively low variability. The high frequencies of similarity levels greater than II for the subjects Nad show that these angles don't reproduce stable. The finding of high movement variability for subject Nad conforms to the other results. It must be assumed that no explicit correlations between Euclidean Distance and coefficient of variation exist.

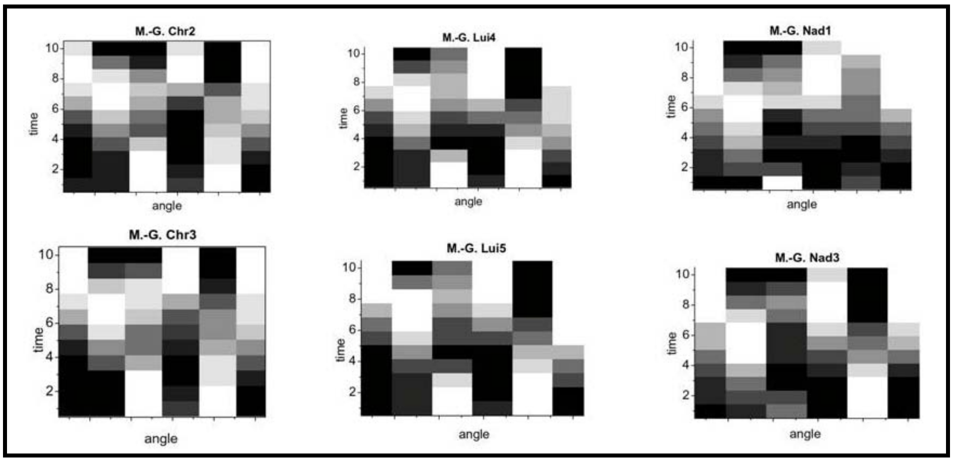

Figure 2. Visualized contour plots of movement pattern of Mae-Geri for 3 athletes, two trials per athlete. White: angle-maximum, black: angle-minimum. From left to right the angles W1(Hip), W2 (Knee), W3 (ankle), W4 (Spine), W5 (PelX), W6 (PelZ) (s. table 1).

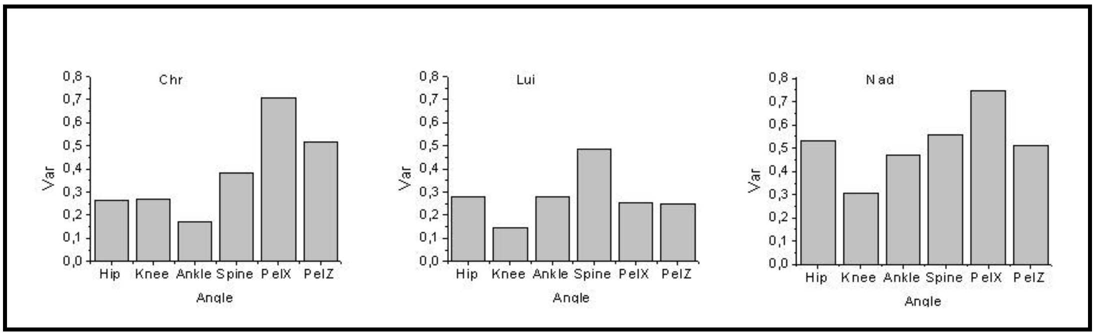

Figure 3. Averaged coefficients of variation for each angle over all executions per subject

DISCUSSION: By means of visualized movement patterns in form of contour plots it was possible to recognize movements with clear differences in the biomechanical characteristics with strong movement artefacts. The discrete angle- time-series could also be used for other analysis: determination of averaged coefficients of variation for each angle over all executions per subject, determination of averaged coefficients of correlation for the single angles over all subjects and calculation of the percentage frequency of the appearance of the similarity levels on the basis of the Euclidean Distance. Generally, the results of these methods are in accordance with the subjective impression of the visual movement patterns. By means of the procedure of the visual movement pattern and the using of the statistical methods some special results recording the similarity of karate kick Mae-Geri could be found. The subjects were able to repeat the movement, with a similar coordination, differently.

Angles which were very stable during repetition (HipX, Knee, PelZ) and angles with a high variability (Ankle, PelX) could be found. From this it is assumed that stable angles are important for the learning process of this movement. The determination of mean coefficients of correlation for the single angles over all subjects confirmed this. The calculation of the percentage frequency of the appearance of the similarity levels on the basis of the Euclidean Distance allows an individual and detailed analysis of the variation of the single angles.

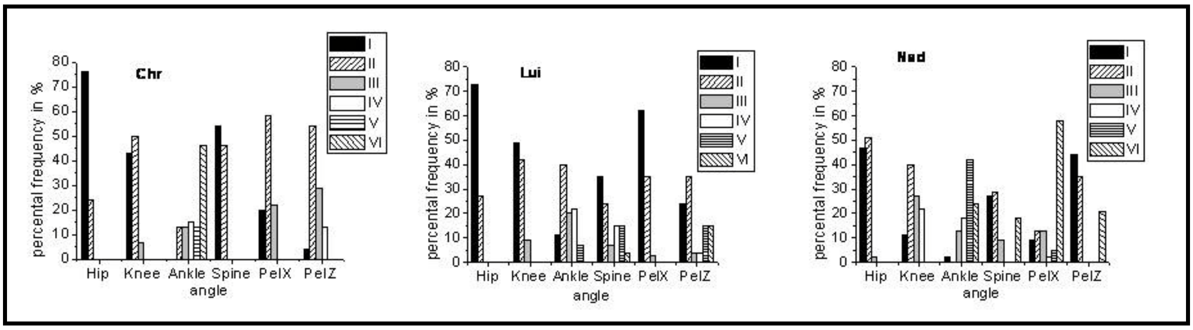

Figure 4. Percentage frequency of the appearance of the similarity levels I – VI for the single angles

CONCLUSION: In summary it can be found that the presented procedure of visualized movement patterns allows a subjective impression of the total movement coordination. Statistical methods of determination of variation provided detailed analyses for the single angles which were used for contour plots for each athlete. From this, expected advice for optimization of the training process can be made.

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ISBS 2010

Oral Session 04

Rapid Movements

New Investigator Award

JUMP KINETICS, BONE HEALTH AND NUTRITION IN ELITE ADOLESCENT FEMALE ATHLETES

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The relationship between physical force capacity (kinetics), nutritional intake, and lower limb bone health was the focus of the present study. 119 adolescent female athletes across four sub-populations, gymnastics, track and field, water polo and non-active controls, completed a series of jump tasks, bone scans, and a three day food diary. Statistical analysis using two-way analysis of variance was used to compare key measures between groups. Significant differences were identified for bone and jump parameters. Stepwise linear regression analysis identified jump kinetics as best able to predict distal tibial trabecular bone density (r^2 = 44.2%, p = 0.000) and bone strength (r^2 = 28.5%, p = 0.000). Athletes engaging in weight-bearing loading appear advantaged in site-specific markers of bone health.

KEYWORDS: bone, loading, training, nutrition, health, sport

INTRODUCTION: For the aspiring adolescent female athlete the competing demands of growth and development and increasing training hours provides a unique challenge when maintaining overall health. The unique nature and demands of a particular sport may also have both short and longer-term health implications. Sports involving considerable loading of the lower limb through high-impact or repetitive ground contacts (e.g. athletics and gymnastics), can illicit both positive (bone building, strength) and negative (injury risk) outcomes (Nichols et al., 2007). Conversely, water-based sports where the body is frequently unloaded during training has differing outcomes of reduced injury risk (positive), but limited potential for bone building (negative).

Nutrition, in particular caloric and calcium intake, also plays a role in bone building, and overall growth and development (Rogol et al., 2000; Nichols et al., 2007). The increased energy demands of training may place additional stress on the nutritional intake of elite athletes. Research highlights that the nutritional habits of elite athletes may be insufficient to support both growth and the rigors of training (Hawley et al., 1995). Insufficient energy intake coupled with high training demands may negatively impact bone health.

The use of jump tasks to measure general strength and power has been widely utilized in applied sports research (e.g. Bradshaw & Le Rossignol, 2004). Through the measurement of kinetics during functional tasks, an understanding of the typical forces experienced in the daily training environment can be obtained. The relationship between nutritional intake, jump performance and bone health is yet to be established, particularly for the elite adolescent female athlete.

The purpose of the present study was to evaluate the relationship between nutritional intake and jump kinetics with site specific markers of bone health in four distinct sub-populations of adolescent females.

METHOD: One hundred and nineteen adolescent female participants from four sub-populations; high-‘impact’ sports (gymnastics and track and field), low-‘impact’ sport (water polo), and a less physically-active control group (<4 hours of physical activity per week outside of their physical education classes) volunteered for the study (refer to Table 1). All participants were injury free at the time of testing. All procedures were approved by the University Ethics Committee and athlete and parental/guardian consent was obtained prior to participation in the study.

Table 1. Participant age, physical descriptors, daily nutritional intake (from 3 day food diary), and weekly physical activity (PA) for all sub-population groups, reported as mean ± standard deviation.

Group	Participants (n)	Age (yrs)	Height (cm)	Mass (kg)	PA (hrs)	Caloric Intake (kCal)[a]	Calcium Intake (mg)[b]
Gymnastics	26	13.7 ± 1.9	146.6 ± 7.8	39.4 ± 7.3	33.2 ± 2.3	1965.6 ± 651.3	992.3 ± 291.3
Track and Field	34	15.9 ± 1.2	168.8 ± 6.8	58.8 ± 7.5	8.4 ± 3.9	2152.6 ± 445.3	996.5 ± 409.5
Water Polo	31	16.2 ± 0.7	172.1 ± 6.1	67.5 ± 8.0	13.0 ± 5.2	2032.3 ± 668.3	810.9 ± 234.3
Controls	28	14.3 ± 1.1	163.9 ± 5.6	58.4 ± 9.3	1.9 ±1.5	2090.1 ± 560.7	1119.6 ± 826.5

Notes: The recommended daily intake for adolescent females aged 12-18 years in Australia and New Zealand is [a]2245 kCal for caloric intake and [b]1300 mg for calcium intake (http://www.nrv.gov.au). The recommendations are an average and don't account for individual variations in physical activity such as those female athletes engaged in elite training.

All of the participants performed a self-administered warm-up prior to the testing. Two warm-up jumps followed by three trials of a counter movement jump (CMJ), a squat jump (SJ), a standing long jump (SLJ), and drop jump (DJ) from a 70cm box were completed. All jumps were performed in the same order. To minimize the effect of fatigue all participants were given approximately 30 seconds recovery time between each jump, and 1-2 minutes between each jump type. During the SJ, a self-selected starting depth was held for 2 s prior to each jump. The participants were instructed to jump as high as possible in the CMJ and SJ trials and jump as high as possible whilst minimizing ground contact time for the DJs.
All of the jumps were completed on two portable, multi-component force plates (Kistler, 9286A, Switzerland) sampling at 1000Hz. Both force plates were covered with Mondo running track surface material to simulate a 'typical' sporting environment surface. Jump height was calculated by using the impulse-momentum method (Linthorne, 2001). Peak force data was normalized to body weight. Jump height and distance was normalized to standing height.
A Peripheral Quantitative Computed Tomography (pQCT; Stratec XCT 2000, Pforzheim, Germany) scanner was used to assess the non-dominant lower limb for each participant at the 4%, 38% and 66% sites of the tibia measured distally. Trabecular area, density and bone strength-strain index (SSI) were assessed at the 4% distal site using the manufacturer's software. Cortical area, density and bone SSI were assessed at the 38% distal site, and muscle and fat area at the 66% distal site.
Means and standard deviations were calculated for all measures across the four sub-populations using Statistical Package for Social Sciences (SPSS, version 17, Chicago, USA). Analysis of covariance (ANCOVA) controlling for limb length with Bonferroni post-hoc analysis was used to identify differences in bone density, area and SSI. Analysis of variance (ANOVA) with Bonferroni post-hoc analysis was used to identify differences between groups for jump kinetics. Pearson product-moment correlations between jump kinetics and bone parameters provided the basis for a stepwise linear regression model to predict bone area, density and SSI from the jump kinetic data.

RESULTS AND DISCUSSION: Table 2 displays the mean tibial bone results. At the 4% distal tibial site, the high-'impact' sport athletes (gymnasts and track & field athletes) had significantly higher trabecular bone density and SSI values ($p<0.001$). This supports the previously reported positive bone building potential of weight-bearing sports (Nichols et al., 2007). In addition, the track and field group displayed significantly higher cortical bone area and SSI ($p<0.001$) at the 38% distal tibial site than the control group. This finding, in the absence of any significant differences in the other sporting groups, would suggest that the type of loading experienced by the track and field athletes (potentially more vertical compression) may be important in cortical bone development.

Table 2. Mean ± standard deviation for distal tibia bone measures of the non-dominant leg.

Group	Tibia Length (mm)	4% Tibia –Trabecular Bone Area (mm^2)	4% Tibia –Trabecular Bone Density (mg/cm^3)	4% Tibia –Trabecular Bone Strength (mm^3)	38% Tibia – Corticol Bone Area (mm^2)	38% Tibia – Corticol Bone Density (mg/cm^3)	38% Tibia – Corticol Bone Strength (mm^3)	66% Tibia Muscle Area (mm^2)	66% Tibia Fat Area (mm^2)
Gymnastics	331.9^{ctw} ± 22.6	472.5 ± 44.4	362.0^{ctw} ± 62.1	2180.9^{cw} ± 301.0	237.5 ± 30.9	1096.7^{ctw} ± 37.9	1180.9 ± 234.3	5549.1^{t} ± 923.7	1363.5^{cw} ± 270.6
Track and Field	378.2^{g} ± 21.4	485.7 ± 75.1	293.2^{cgw} ± 54.6	2235.6^{cw} ± 557.7	293.9^{cw} ± 35.7	1140.2^{g} ± 26.1	1643.1^{c} ± 304.5	6606.7^{gw} ± 806.9	2048.4^{cw} ± 499.2
Water Polo	384.3^{cg} ± 22.4	468.3^{c} ± 56.0	231.2^{gt} ± 41.7	1831.6^{gt} ± 423.8	271.7^{t} ± 25.5	1151.2^{g} ± 23.4	1563.9 ± 231.8	5933.9^{t} ± 849.7	3054.4^{gt} ± 1412.9
Controls	366.1^{gw} ± 35.3	501.5^{w} ± 53.4	241.4^{gt} ± 36.4	1665.7^{gt} ± 498.4	242.8^{t} ± 35.6	1135.7^{g} ± 28.4	1322.6^{t} ± 303.4	5971.9 ± 717.6	2637.3^{gt} ± 759.9

Notes: Statistically significant with; c: controls, g: gymnasts, t: track and field, w: water polo. ($p<0.05$ following Bonferroni adjustment for multiple comparisons) All bone measures were covaried for tibial length. 'Estimated' mean values following covariance not reported.

Jump kinetic and performance data is presented in Table 3. Overall, the gymnasts displayed significantly greater peak forces than the water polo (all jumps, $p<0.005$), control (SJ and CMJ, $p<0.001$) and track and field athletes (SJ and SLJ, $p<0.001$). Track and field athletes and gymnasts had significantly better jump performance than the controls and water polo athletes across all tests ($p<0.001$) except for the DJ where only gymnasts showed a significant difference in jump height. Water polo athletes had the lowest peak forces across all groups with the controls displaying the lowest jump heights and distances across all tests. Gymnasts had a significantly shorter contact time (0.240s) for the DJ test than the track and field (0.300s), control (0.333s) and water polo (0.363s) groups.

Table 3. Peak ground reaction forces and jump height/distance as a percentage of standing height (m) for the four jump tasks for all groups, reported as means ±standard deviation. All force measures are vertical with the exception of the SLJ which is a horizontal force.

Group	CMJ Force (BW)	CMJ Height (%)	SJ Force (BW)	SJ Height (%)	SLJ Force (BW)	SLJ Distance (%)	DJ Force (BW)	DJ Height (%)
Gymnastics	2.5^{w} ± 0.3	0.15^{cw} ± 0.02	2.8^{ctw} ± 0.3	0.14^{cw} ± 0.02	1.2^{ctw} ± 0.1	1.20^{cw} ± 0.10	9.2^{w} ± 1.4	0.10^{cw} ± 0.03
Track and Field	2.3 ± 0.2	0.14^{cw} ± 0.03	2.5^{gw} ± 0.4	0.14^{cw} ± 0.03	0.9^{g} ± 0.1	1.16^{cw} ± 0.11	8.2 ± 1.8	0.08 ± 0.03
Water Polo	2.2^{g} ± 0.2	0.12^{cgt} ± 0.02	2.2^{gt} ± 0.3	0.11^{cgt} ± 0.02	0.7^{g} ± 0.1	0.92^{gt} ± 0.11	7.8^{cg} ± 1.4	0.07^{g} ± 0.03
Controls	2.3 ± 0.3	0.10^{gtw} ± 0.02	2.3^{g} ± 0.2	0.09^{gtw} ± 0.02	0.8^{g} ±0.1	0.90^{gt} ± 0.12	9.3^{w} ± 1.7	0.06^{g} ± 0.03

Notes: Statistically significant with; c: control group, g: gymnastics group, t: track and field group, w: water polo group ($p<0.05$ following Bonferroni adjustment for multiple comparisons).

Regression analysis showed reasonable predictability of trabecular bone density and strength at the 4% distal tibia site with jump kinematics and kinetics able to explain 43.5% (p=0.000) and 28.5% (p=0.000) of the variance in bone scores respectively. Predictability of cortical bone variance using jump kinetics was not as high with only 14.1% (p=0.000) of cortical bone density at the 38% distal tibia site explained by SLJ and SJ peak force. Caloric and calcium intake were not significantly related to any of the bone parameters.

Table 4. Regression analysis using jump kinetics to predict differences in bone area, density and SSI.

Dependent Variable	Model	R^2	P	SEE%	Formula & independent variables
4% Tibia – Trabecular Bone	Area (mm^2)	0.037	0.029	12.2%	451.29 + 388.32 x DJ Height
	Density (mg/cm^3)	0.435	0.000	18.8%	7.94 + 216.51 x SLJ Peak Force + 8.66 x DJ Peak Force
	SSI (mm^3)	0.285	0.000	22.0%	200.84 + 930.89 x SLJ Distance + 3764.27 x DJ Jump Height + 58.13 x DJ Peak Force
38% Tibia – Corticol Bone	Area (mm^2)	0.048	0.015	14.5%	321.17 – 23.55 x SJ Peak Force
	Density (mg/cm^3)	0.141	0.000	2.9%	1216.62 – 19.91 x SJ Peak Force -38.93 x SLJ Peak Force
	SSI (mm^3)	0.080	0.013	21.6%	1907.32 – 504.00 x SLJ Peak Force

CONCLUSION: Elite adolescent female athletes involved in 'high-impact' sports (gymnastics and track & field) showed greater trabecular bone density and bone strength as well as increased jumping ability and force production capability compared with 'low-impact' (water polo) sporting and control groups. The ability of jump kinetics to predict differences in trabecular density and bone strength at the distal tibia among athletic populations suggests that the ability to produce and routinely deal with higher forces is important for bone development. The type of loading experienced during sports participation appears important for cortical bone development. The repetitive, more vertical and compressive nature of loading on relatively hard surfaces experienced in track and field differs from the repetitive, more rotational nature of loading on predominantly sprung surfaces experienced by gymnasts. This varies with the loading demands of a non weight-bearing sport such as water polo. These demands, as suggested by the differences in kinetic capacity between the groups, appear to have an impact on bone health. However, the exact relationship between force capacity and bone health may be more evident during longitudinal assessment.

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HAMSTRINGS, QUADRICEPS, AND GLUTEAL MUSCLE ACTIVATION DURING RESISTANCE TRAINING EXERCISES

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This study evaluated hamstrings, quadriceps, and gluteal muscles activation during the back squat, deadlift, step-up, and lunge. Root mean square electromyographical data were analyzed for the eccentric and concentric phases during each exercise. Subjects included 16 women who performed 2 repetitions of each of the exercises at a 6 repetition maximum load. A repeated measures ANOVA revealed significant main effects ($P \leq 0.05$) representing differences in muscle activation between the resistance training exercises all muscle groups ($P \leq 0.05$) except for the rectus femoris, during the concentric phase ($P=0.22$). Based on these results, resistance training exercises can be prescribed based on how they best train the desired musculature.

KEYWORDS: women, strength training,

INTRODUCTION: The quadriceps, hamstrings, and gluteal muscles are important in sport performance and injury prevention. The quadriceps group raises as well as controls the descent of the body's center of mass during athletic movements, such as running and jumping (Neumann, 2010). However, some evidence indicates that strong quadriceps forces cause anterior translation of the tibia and increase the risk of injury to the anterior cruciate ligament (ACL) (Baratta et al., 1988). Hamstring training may reduce hamstrings inhibition, hamstrings to quadriceps imbalance, and ACL injuries (Baratta et al., 1988).

Training the quadriceps disproportionately to the hamstrings may inhibit hamstrings co-activation, reduce joint stability, increase anterior tibial translation in response to strong quadriceps forces (Hewett et al., 2001), and potentially increase the incidence of hamstring strains (Yamamoto, 1993). Understanding hamstrings to quadriceps activation ratios is potentially important for performance and injury prevention during a variety of athletic movements such as jump landings, single leg support, and cutting maneuvers (Yamamoto, 1993). Such athletic movements also place a large demand on the hip musculature.

The gluteus medius and maximus have attachments on both the femur and pelvis, and therefore contribute not only to movements of the lower extremity, but also to pelvic and trunk stabilization. Insufficient strength or recruitment of these muscles may contribute to poor core stability, as well as to malalignment of the lower extremity. Training hip muscles increases lower extremity alignment, improves landing technique, and decreases risk of ACL injury by reducing valgus force to the knee (Myer et al., 2008).

Few studies have evaluated lower body muscles activation during a variety of resistance training exercises. Studies have assessed unloaded single leg squats, lunges and a 20.32 cm step-up exercise (Bourdreau et al., 2009) and hamstrings and quadriceps activation during a variety of loaded resistance training exercises (Ebben, 2009; Ebben et al., 2009). At present, no study has evaluated the activation of hip and knee muscles groups for several lower body resistance training exercises that incorporate hip and knee extension. Therefore, the purpose of this study was to quantify muscle activation of the quadriceps, hamstrings, and gluteal muscles during the back squat, deadlift, step-up, and lunge.

METHODS: Subjects included 16 women (mean ± SD; age = 21.19 ± 2.17 years; height = 169.39 ± 7.54 cm; body mass = 66.08 ± 9.91 kg) who participated in either NCAA Division I or club or intramural sports and lower body resistance training. All subjects provided informed consent and the university's internal review board approved the study.

Subjects attended one pre-test habituation session and one testing session. Prior to each, subjects participated in a standardized general and dynamic warm up. During the pre-test habituation session, subjects were familiarized with and performed their 6 repetition maximum (6 RM) for the back squat, deadlift, step-up, and forward lunge. All exercises were performed according to the methods previously described (Earle & Baechle, 2000) with the exception that the step-up began on top of the box so that all exercises started with the eccentric phase and ended with the concentric phase.

Following the 6 RM testing, subjects were familiarized with 4 maximum voluntary isometric contraction (MVIC) tests for the hamstrings, quadriceps, gluteus medius, and gluteus maximus.

Approximately 1 week after the pre-test habituation session, subjects returned for the testing session. During this session, subjects performed MVICs for the hamstrings, quadriceps, gluteus medius, and gluteus maximus with contractions held for 6 seconds each. Subjects then were tested by performing 2 full range of motion repetitions of their previously determined 6 RM loads, for each of the test exercises. Randomization of the exercises, limited repetitions, and 5 minutes of recovery were provided between MVICs as well as each test exercise.

Surface electromyography (EMG) was used to quantify muscle activation using a fixed shielded cabled, telemetered EMG system (Myomonitor IV, DelSys Inc. Boston, MA, USA). Data were recorded at sample rate of 1024 Hz using bipolar surface electrodes with 1 x 10 mm 99.9% Ag conductors, and an inter-electrode distance of 10 mm. Electrodes were placed on the longitudinal axis of the medial and lateral hamstrings (MH and LH, respectively) the rectus femoris (RF), the vastus lateralis and medialis (VL and ML, respectively), and the gluteus medius and maximus (GMD and GMX, respectively). A common reference electrode was placed on the lateral malleolus. Electrode placement was chosen in order to assess uni-articular and bi-articular knee extensor and flexor muscles, as well as hip abductors and extensors. Additionally, an electric goniometer was placed on the lateral aspect of the right knee in order to distinguish between the eccentric and concentric phases of the test exercises. Skin preparation included shaving, abrasion and cleansing with alcohol. Elastic tape was applied to ensure electrode placement and provide strain relief for the electrode cables. Surface electrodes were connected to an amplifier and streamed continuously through an analog to digital converter (DelSys Inc. Boston, MA, USA) to an IBM-compatible notebook computer.

All data were filtered with a 10-450 Hz band pass filter, saved, and analyzed with the use of software (EMGworks 3.1, DelSys Inc., Boston, MA, USA). The input impedance was 1015 Ohms and the common mode rejection ratio was >80 dB. Raw data were acquired and processed using root mean square (RMS) EMG with a moving window of 125 ms. Electromyographic data were analyzed for seconds 2-3 of the MVICs, and for eccentric and concentric phases for each of the 4 test exercises using the average of both test repetitions. All RMS EMG values for each muscle were normalized to the average RMS EMG of the 2 trials of the MVIC.

Data were evaluated with a repeated measures ANOVA to test main effects of RMS EMG for each muscle assessed. Bonferroni adjusted post hoc analyses were used to assess the specific differences in muscle activation between the resistance training exercises. The *a priori* alpha level was set at $P \leq 0.05$ and all data are expressed as means ± SD.

RESULTS: Significant main effects representing differences in muscle activation between the resistance training exercise were found for all muscle groups ($P \leq 0.05$) except for the RF, during the concentric phase ($P = 0.22$). Table 1-7 shows the differences in muscle activation between the exercises. Subjects' mean squat, deadlift, lunge and step-up estimated 1 RMs were 88.98, 83.66, 67.38 and 38.48 kg, respectively.

Table 1. RMS EMG data for the lateral hamstring (LH) during the eccentric (ECC) and concentric (CON) phases of the 4 study exercises. (N=14)

	Deadlift	Lunge	Step-Up	Squat
LH ECC	0.55 ± 0.51[a]	0.38 ± 0.27[b]	0.35 ± 0.27[b]	0.28 ± 0.20[b]
	Deadlift	Step-Up	Lunge	Squat
LH CON	1.29 ± 0.72[a]	0.87 ± 0.54[b]	0.86 ± 0.66[b]	0.62 ± 0.40[b]

a = significantly different than all other exercises (p ≤ 0.05); b = significantly different than DL (p ≤ 0.05).

Table 2. RMS EMG data for the medial hamstring (MH) during the eccentric (ECC) and concentric (CON) phases of the 4 study exercises. (N=14)

	Deadlift	Lunge	Step-Up	Squat
MH ECC	0.43 ± 0.31[a]	0.41 ± 0.33[a]	0.31 ± 0.24[b]	0.24 ± 0.21[b]
	Deadlift	Step-Up	Lunge	Squat
MH CON	0.90 ± 0.41[c]	0.59 ± 0.36[d]	0.57 ± 0.49[d]	0.49 ± 0.29[d]

a = significantly different than SU and S (p ≤ 0.05); b = significantly different than DL and L (p ≤ 0.05); c = significantly different than all other exercises (p ≤ 0.01); d = significantly different than DL (p ≤ 0.05)

Table 3. RMS EMG data for the rectus femoris (RF) during the eccentric (ECC) phase of the 4 study exercises. (N=14)

	Squat	Lunge	Step-Up	Deadlift
RF ECC	0.81 ± 0.35[a]	0.66 ± 0.38[b]	0.63 ± 0.27[c]	0.54 ± 0.49[d]

a = significantly different than DL (p ≤ 0.05); b = significantly different than S and DL (p ≤ 0.01); c = significantly different than DL and SU (p ≤ 0.01); d = significantly different than S, L, SU (p ≤ 0.01);

Table 4. RMS EMG data for the vastus lateralis (VL) during the eccentric (ECC) and concentric (CON) phases of the 4 study exercises. (N=14)

	Lunge	Squat	Step-Up	Deadlift
VL ECC	0.98 ± 0.47[a]	0.94 ± 0.40[a]	0.90 ± 0.41[a]	0.61 ± 0.28[b]
VL CON	1.37 ± 0.59[a]	1.33 ± 0.68[a]	1.31 ± 0.82[a]	0.63 ± 0.32[b]

a = significantly different than DL (p ≤ 0.05); b = significantly different than all other exercises (p ≤ 0.05)

Table 5. RMS EMG data for the vastus medialis (VM) during the eccentric (ECC) and concentric (CON) phases of the 4 study exercises. (N=14)

	Lunge	Step-Up	Squat	Deadlift
VM ECC	1.27 ± 0.54[a]	1.22 ± 0.61[a]	1.17 ± 0.49[a]	0.78 ± 0.34[b]
VM CON	1.77 ± 0.63[a]	1.73 ± 0.94[a]	1.49 ± 0.54	1.32 ± 0.68[c]

a = significantly different than the DL (p ≤ 0.001); b = significantly different than all other exercises (p ≤ 0.01); c = significantly different than SU and L (p ≤ 0.01)

Table 6. RMS EMG data for the gluteus medius (GMD) during the eccentric (ECC) and concentric (CON) phases of the 4 study exercises. (N=14)

	Step-Up	Lunge	Deadlift	Squat
GMD ECC	0.56 ± 0.27[a]	0.55 ± 0.30[a]	0.25 ± 0.09[b]	0.23 ± 0.11[b]
GMD CON	0.85 ± 0.27[a]	0.84 ± 0.35[a]	0.56 ± 0.34[b]	0.38 ± 0.15[b]

a = significantly different than S and DL (p ≤ 0.001); b = significantly different than SU and L (p ≤ 0.001)

Table 7. RMS EMG data for the gluteus maximus (GMX) during the eccentric (ECC) and concentric (CON) phases of the 4 study exercises. (N=14)

	Lunge	Step-Up	Deadlift	Squat
GMX ECC	0.95 ± 0.45[a]	0.87 ± 0.31	0.76 ± 0.36[b]	0.62 ± 0.34[b]
	Step Up	Lunge	Deadlift	Squat
GMX CON	1.99 ± 0.91[c]	1.88 ± 0.69[c]	1.79 ± 0.88[c]	1.18 ± 0.50[d]

a = significantly different than DL and S ($p \leq 0.05$); b = significantly different than L ($p \leq 0.05$); c = significantly different than S ($p \leq 0.05$); d = significantly different than all other exercises ($p \leq 0.05$)

DISCUSSION: This study is the first to assess GMD and GMX activation during a variety of lower body resistance training exercises and further investigates hamstrings and quadriceps muscle activation during eccentric and concentric phases of a variety exercises. Results indicate that exercises such as the step-up and lunge are best for GMD and GMX activation, potentially due to the unilateral nature of these exercises. As a result, exercises such as these should be included in a training program for sports that require hip extension and abduction for dynamic performance or stabilization. Previous research demonstrated that single leg squat produced more GMX and GMD than lunges or 20.32 cm step-up exercises (Bourdreau et al., 2009). However, these exercises were performed without any added resistance (Bordreau et al., 2009). Of the exercises assessed in the present study, the deadlift appears the be the best MH and LH activator, producing 55 and 43 percent of the MVIC, respectively, during the eccentric phase, and 129 and 90 percent of the MVIC, respectively, during the eccentric phase. Previous research demonstrated biceps femoris activation of approximately 55 percent of the MVIC during the deadlift (Ebben et al., 2010). Results of the present study demonstrate mean rectus femoris and vastus lateralis activation ranged from highest to lowest during squat, lunge, step-up and deadlift. This finding is identical to previous research (Ebben et al., 2010).

VM activation has not been previously assessed during a variety of lower body resistance training exercises. In the present study, VM activation was highest during the lunge and step-up while the deadlift produced significantly lower levels of VM activation.

CONCLUSION: Results of this study demonstrated that the step-up and lunge are the best exercises for activating the GMD, GMX, and VM. The squat best activates the RF and the lunge, squat, and step-up are equally effective at activating the VL. The deadlift is the best exerciser for activating the MH and LH. Resistance training exercises should be prescribed based on how they best train the desired musculature.

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ANTAGONIST CONDITIONING CONTRACTIONS IMPAIR AGONIST FUNCTIONING

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This study assessed the effect of antagonist conditioning contractions (ACC) on the subsequent force and electromyography of an agonist. Twelve subjects performed isokinetic elbow flexion on a dynamometer in 4 test conditions including a baseline condition without, and 1, 3, and 6 second durations of, isometric triceps extension prior to elbow flexion. Average peak torque (T), peak torque/body weight (T:BW), average power (P), and rate of torque development (RTD) were assessed. Electromyographic data were obtained from elbow extensors and flexors. A repeated measures ANOVA with post hoc analysis demonstrated that T, T:BW, P, and RTD were higher in the baseline, compared to the ACC conditions ($P \leq 0.05$), and appears to be due to higher brachioradialis activation in the baseline condition in compared to some ACC conditions ($P \leq 0.05$).

KEY WORDS: reversal of antagonists, successive induction, Golgi tendon organ, superset

INTRODUCTION: The activation of the antagonist in order to potentiate the agonist muscle group has been referred to as the successive induction and is thought to stimulate the Golgi tendon organ (GTO) (Kroll, 1972). The stimulation of the GTO in an antagonist, may inhibit its activation and stimulate the activation in a subsequently contracted agonist. In addition to successive induction, other similar terms for this process include the reversal of antagonists, which is present in the rapid transitions between antagonist and agonist muscle groups during movements such as walking, running, and rowing (Voss et al., 1985). Skilled athletes appear to be able to reduce antagonist co-activation as an adaptation that allows them to produce greater agonist force (Bazzucchi et al., 2008).
In addition to successive induction and the reversal of antagonists, a small body of literature defines these phenomena as antagonist conditioning contractions (ACC). Research examining the role of ACC on agonist force demonstrates higher rates of force development but not higher peak force, and no evidence of increased muscle activation (Gabriel et al., 2001; Grabiner et al., 1994; Kamimura et al., 2009). Variations exist in the magnitude and duration of the antagonist activation which may affect the ergogenic potential of this stimulus. Furthermore, some evidence demonstrates that stimulating a muscle with a maximal or near maximal activation may potentiate rather than inhibit it (Hodgson et al., 2005) and efforts to reduce the activation of the antagonist via a fatiguing stimulus resulted in its potentiation and subsequent impairment of agonist functioning (Maynard and Ebben, 2003). Finally, some evidence indicates that the antagonist inhibition may last only 1 second (Chalmers, 2004). Thus, the challenge seems to be to activate the antagonist enough to stimulate the GTO, while not potentiating it, and to take advantage of the antagonist inhibition before it decays. Therefore, the purpose of this study was to assess the duration of ACC on subsequent performance and activation of the agonist.

METHODS: Twelve men (mean ± SD: age = 21.08 ± 1.80 years; height = 183.54 ± 8.88 cm; body mass = 83.39 ± 9.35 kg; frequency of resistance training = 3.25 ± 0.75 days/week) volunteered to serve as subjects for the study. Subjects signed an informed consent form prior to participating in the study which was approved by the Institutional Review Board.
Subjects performed a general and dynamic warm up prior to the study. Subjects then performed a task specific warm up on the dynamometer (System 4, Biodex Inc., Shirley, NY)

consisting of 2 sets of 1 repetition of isometric (ISOM) elbow extension at 75 and 90% of their self perceived maximum ability and 2 sets of 2 repetitions of isokinetic (ISOK) elbow flexion at 75 and 90% of their self perceived maximum ability. Subjects then rested for 5 minutes and performed the test sets.

Subjects performed 4 test sets in random order with 5 minutes of recovery between the sets. Test sets included a baseline ISOK elbow flexion set without a preceding ACC set, and 3 other test sets of ISOK elbow flexion following 1, 3, and 6 second durations of ISOM ACC elbow extension. For each test set, kinetic and muscle activation data were collected using dynamometry and electromyography (EMG), respectively.

For the test sets, subjects were positioned in the dynamometer according to manufacturer specifications. The system was calibrated with the system software. The right elbow was positioned goniometrically at a starting position of 10°of elbow flexion. Isometric ACC and ISOK elbow flexion was performed from this starting point. Isokinetic elbow flexion was performed at 60° per second through a range of motion of 120° of elbow flexion. The order of test sets was randomized with 5 minutes of recovery between tests to reduce order and fatigue effects.

Torque curves for each subject were analyzed using manufacturer's software. Data were sampled for the entire range of motion of the ISOK test sets. Peak torque (T), torque to body weight ratio (T:BW), power (P), and rate of torque development (RTD) were calculated as the average of the two repetitions of each ISOK test sets. Rate of torque development was calculated for the first 300 ms of each test exercise and normalized to a second.

Electromyography was used to quantify muscle activity using a telemetered EMG system (Myomonitor IV, DelSys Inc. Boston, MA, USA). The input impedance was 10^{15} Ohms with a common mode rejection ratio of >80 dB. Electroymyographic data from the biceps brachii (BB) and brachioradialis (BR), were used to assess the agonist elbow flexors, consistent with previous work assessing ACC (Holt et al., 1969). Electromyographic data were also recorded from the triceps brachii-long head (TB-Long) and triceps brachii-lateral head (TB-Lateral) in order to assess muscle activation of the antagonist during all test conditions. Data were recorded at a sampling rate of 1024 Hz using rectangular shaped (19.8 mm wide and 35 mm long) bipolar surface electrodes with 1 x 10 mm 99.9% Ag conductors, and an inter-conductor distance of 10 mm. A common reference electrode was placed on the lateral malleolus of the right leg. Skin preparation included shaving hair if necessary, abrasion, and cleaning the surface with alcohol. Elastic tape was applied to secure electrode placement in order to minimize motion artifact and to provide strain relief for the electrode cables. Surface electrodes were connected to an amplifier and streamed continuously through an analog to digital converter (DelSys Inc. Boston, MA, USA) to an IBM-compatible notebook computer.

All data were filtered with a 10-450 Hz band pass filter, saved, and analyzed with the use of software (EMGworks 3.1, DelSys Inc., Boston, MA, USA). Root mean square signal processing was used and data were calculated using a 125 ms moving window. Root mean squared EMG data were analyzed for the muscle burst for ISOM ACC and the ISOK elbow flexion tests. Burst onset and offset was determined as the points at which the RMS EMG values initially exceeded and eventually fell below 150 percent of baseline EMG values for each muscle burst. Data were averaged for the two trials and normalized to a resting value for each muscle assessed.

Data were analyzed with SPSS 16.0 using a repeated measures ANOVA and Bonferroni adjusted pairwise comparisons to identify the specific differences in T, T:BW, P, and RTD and RMS EMG for each muscle assessed between the test conditions. The criterion for significance was set at $P \leq 0.05$. Effect sizes and power were determined η_p^2 and d, respectively.

RESULTS: Statistical analysis revealed a significant main effect for T ($P = 0.001$, $\eta_p^2 = 0.64$, $d = 1.00$), T:BW ($P \leq 0.05$, $\eta_p^2 = 0.34$, $d = 0.92$), P ($P = 0.003$, $\eta_p^2 = 0.34$, $d = 0.92$) and RTD ($P \leq 0.001$, $\eta_p^2 = 0.58$, $d = 1.00$) demonstrating differences between the test conditions for these variables. Table 1 shows the specific differences for these variables based on post hoc analysis. Statistical analysis of RMS EMG data revealed significant main effects for BB

(P = 0.04, η_p^2 = 0.19, d = 0.54), BR (P = 0.04, η_p^2 = 0.23, d = 0.68), TB-long (P = 0.33), and TB-lateral (P = 0.72). Table 2 shows the specific differences for these variables based on post hoc analysis.

Table 1. Peak torque (T), torque to body weight ratio (T:BW), power (P), and rate of torque development (RTD) for isokinetic elbow flexion in baseline and 3 ACC conditions.

Condition	T(N)*	T:BW*	P(w)**	RTD(N·sec^{-1})*
Baseline (No ACC)	65.30 ± 12.26	81.06 ± 11.22	46.12 ± 11.72	154.60 ± 22.98
1 sec ACC	60.06 ± 10.39	73.88 ± 9.19	39.30 ± 8.73	130.49 ± 17.14
3 sec ACC	60.18 ± 10.19	73.67 ± 8.55	40.37 ± 8.66	124.14 ± 24.34
6 sec ACC	59.61 ± 10.33	73.39 ± 9.67	38.38 ± 9.29	132.73 ± 21.17

*Baseline condition is significantly different than all ACC conditions ($p \le 0.01$).
**Baseline condition is significantly different than all ACC conditions ($p \le 0.05$).

Table 2. Muscle activation (millivolts) expressed as RMS EMG normalized to resting values for biceps brachii (BB), brachioradialis (BR), triceps brachii-long head (TB-long) and triceps brachii-lateral head (TB-Lateral) in the baseline and 3 ACC conditions.

	Agonist		Antagonist	
Condition	BB*	BR**	TB-Long	TB-Lateral
Baseline (No ACC)	0.803 ± 0.24	0.525 ± 0.11	0.032 ± 0.01	0.050 ± 0.02
1 sec ACC	0.816 ± 0.26	0.497 ± 0.09	0.031 ± 0.01	0.051 ± 0.02
3 sec ACC	0.785 ± 0.24	0.517 ± 0.11	0.033 ± 0.02	0.051 ± 0.02
6 sec ACC	0.837 ± 0.28	0.530 ± 0.12	0.033 ± 0.01	0.052 ± 0.03

*3 second ACC is significantly different than 6 second ACC condition ($p \le 0.05$).
**Baseline condition is significantly different than 1 second ACC; 1 second ACC is significantly different than 6 second ACC ($p \le 0.05$).

DISCUSSION: This study demonstrates that maximal isometric ACC impaired subsequent agonist performance for all variables assessed, regardless of the duration of the ACC. This impairment appears to be due to higher levels of BR activation in the baseline condition compared to some of the conditions, following the ACC. No evidence of the inhibition of the antagonist was found in 1, 3, and 6 second durations of the ACC. These results raise questions about the effectiveness of activating the antagonist in order to augment performance in a subsequently activated agonist.

Results of the present study differ from previous research examining the effect of ACC which demonstrated increased rate of force development (Gabriel et al., 2001; Grabiner et al., 1994; Kamimura et al., 2009), but not force (Grabiner et al., 1994; Kamimura et al., 2009) or work (Grabiner et al., 1994).

Most studies assessing ACC failed to find any increase in EMG of the agonist (Gabriel et al., 2001; Holt et al., 1969; Kamimura et al., 2009) demonstrating that either EMG was unable to detect, or another mechanism was responsible for, the increase rate of force development demonstrated in these studies. In contrast, results of the present study show some differences in muscle activation of the agonist among the ACC conditions. However, despite the differences, the data demonstrate the agonist's kinetic performance is greatest in the baseline condition. The present study also shows that there is no antagonist inhibition as assessed by EMG, regardless of the duration of the ACC. This may potentially be due to the approximate 10 second duration between the cessation of the ACC and the onset of the agonist contraction. Previous reports suggest that the inhibition of the antagonist may last only one second (Chalmers, 2004). If so, this limitation reduces the practical benefit of ACC for resistance training, since functional benefits of reversal of antagonists may be present during a variety of functional movement that quickly transition between antagonist and agonist such as walking or running (Voss et al., 1985) or the potential for chronic adaptation in skilled performers (Bazzucchi et al., 2008).

CONCLUSION: This study demonstrated that maximal short term ACC do not enhance, and appear to impair, kinetic performance and activation of prime movers in some conditions. No evidence of inhibition of the antagonist was found. These findings provide evidence that the use of ACC to enhance agonist performance may not be effective. The activation of the antagonist shortly before the activation of an agonist muscle group during resistance training may have not be beneficial and may possibly be detrimental. Thus, agonist/antagonist, push pull, and compound set resistance training strategies should be avoided in cases where maximum force development is desired.

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ACTIVATION AND CONTRIBUTION OF TRUNK AND LEG MUSCULATURE TO FORCE PRODUCTION DURING ON-WATER SPRINT KAYAK PERFORMANCE.

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Velocity production during sprint kayaking has been shown to be dependent on the magnitude of forces produced during the stroke cycle. However, while the importance of the upper body in force production has been promoted by previous research, the importance of the trunk and lower body are yet to be established. Eight international level paddlers completed 5 on-water sprint trials during which paddle force and trunk and leg muscle activations were recorded. Significant correlations ($p<0.05$) were identified between peak force and peak contralateral rectus abdominus activation, while the left external oblique demonstrated significant correlations ($p<0.05$) with peak and mean force during both left and right paddle strokes. Results indentify that the lower trunk plays an important role in force production and therefore sprint performance.

KEYWORDS: Electromyography, Kayaking, Force production

INTRODUCTION: Propulsion during sprint kayaking is dependent on the magnitude of the force that can be developed during the paddle stroke (Mononen *et al*., 1994; Mononen and Viitasalo, 1995). Petrone *et al*. (2006) recorded peak force values ranging between 253 N and 465 N, while other researchers have commonly recorded peak force values above 200 N (Mononen *et al*., 1994; Mononen and Viitasalo, 1995). Despite the clear importance of force, the majority of previous technique research has focused upon the positioning and motions of the upper limbs and as such suggested these to be key in kayak propulsion. However, the magnitude of the forces required during a single paddle stroke and the high stroke rates exhibited (60-70 spm) would result in extreme demands being placed upon the small muscles of the upper limbs. Therefore investigations of other muscles with possible contributions to propulsion are required. Furthermore, Lovell and Lauder (2001) identified, through maximal on-ergometer testing, that bilateral strength imbalances are prevalent in kayakers, predisposing athletes to injury. Whilst Aitken and Neal (1992) and Mononen and Viitasalo (1995) further identified bilateral asymmetry in force production during on-water paddling.

Asymmetries although well established within the arena of force production are yet to be established in the activation levels of muscles during kayaking. Previous work has clearly overlooked this factor with the only on-water study measuring muscular activation being collected from the musculature on a single side athlete (Fleming *et al*. 2007). A bilateral analysis has been conducted on-ergometer (Logan and Holt, 1985). However, only basic sequencing data has been presented, with theories put forward sugg*esting* that the role of the upper limbs is to ensure correct orientation of the paddle whilst large unspecified muscles provide the majority of the propulsive work. Consequently, the lack of empirical findings limits the validity of these claims until such research is conducted to corroborate such propositions. Resultantly, this paper will investigate the activations of the large muscles of the legs and trunk, in accordance with the teachings of Kemecsey (1986) and the on-ergometer findings of Logan and Holt (1985), focusing on their contributions to force production. Furthermore, this analysis will be deconstructed into left and right paddle strokes in accordance with the asymmetrical findings of previous research.

METHOD: Eight male (n=6) and female (n=2) international level paddlers participated after completing informed consent and health questionnaires. Subjects were prepared with Blue Sensor passive surface electrodes over the muscle belly of the latissimus dorsi (LD), rectus abdominus (RA), external oblique (EXO), rectus femoris (RF), biceps femoris (BF) and gastrocnemius (G) on both left (L) and right (R) sides. Each subject used their own paddle with bespoke strain gauges (Sperlich and Sperlich, Germany) mounted perpendicular to the blade. Prior to testing, maximal voluntary contractions (MVC) and force calibration were conducted to normalise all data. Each subject completed 5 trials over a 75m distance, comprising of a 50m acceleration sector, 5m calibrated volume and 20m run off. Data was analysed using Myodat v6.47 and Sportlogger software, from which peak RMS EMG for each muscle, and peak and mean force were calculated for each stroke within the 5m calibrated volume. Paired samples t-tests were conducted to compare peak and mean values during left and right strokes, while correlations and linear regressions between peak activation and force were used to determine relationships between propulsive force and muscle activation.

RESULTS: Comparison of force production identified a significantly higher mean force (MF) during the left stroke (Left: 239.9 N; Right: 208.3 N, $p<0.05$), while peak force (PF) and loading rate (LR) displayed no difference (Table 1).

Table 1. Intra stroke performance descriptives of paddle force.

	Left Stroke	Right Stroke
Mean Force (N)	239.9±13.6*	208.3±17.4*
Peak Force (N)	365.1±24.7	343.6±43.1
Loading Rate ($N.s^{-1}$)	2062.3±369.7	1880.3±347.5

*** denotes significant difference (p<0.05) between left and right strokes.**

Significant differences in peak activation between left and right stroke were identified in the left and right latissimus dorsi alone (Figure 1). Peak left (r = 0.680) and right (r = 0.855) rectus abdominus activation during the left stroke exhibited significant positive relationships with mean force (p<0.05). The right rectus abdominus also displayed a significant predictive relationship with mean force (R^2 = 0.731, $p<0.05$) and a significant positive relationship (r = 0.651, $p<0.05$) with peak force production during the left stroke. The left external oblique displayed significant positive relationships with both mean (r = 0.801, $p<0.05$) and peak (r = 0.798, $p<0.05$) force during the left paddle stroke. The right stroke was characterised by significant positive relationships between mean and peak force and the left external oblique (MF: r = 0.663, PF: r = 0.643, $p<0.05$) and rectus abdominus (MF: r = 0.944; PF: r = 0.955, $p<0.05$) (Table 2).

DISCUSSION: The significant difference identified between left and right paddle strokes within mean force corroborates the findings of previous researchers (Aitken and Neal, 1992; Mononen and Viitasalo, 1995; and Lovell and Lauder, 2001).Thus, indicating a clear dependence on the left stroke in this group of subjects; this raises the question of handedness, although this data was not recorded. However, as a propulsive mechanism this would only affect lateral motion of the kayak if the positioning of the blade does not apply the force in the appropriate direction. Furthermore, the clear activation of the rectus abdominus and rectus femoris holding the lower trunk and ipsilateral leg in a braced position would aid in ensuring that the force be directed longitudinally down the kayak in the intended direction of travel. As such, the variation between sides although significant is not necessarily detrimental to performance; this imbalance may, however, predispose athletes to injury (Lovell and Lauder, 2001).

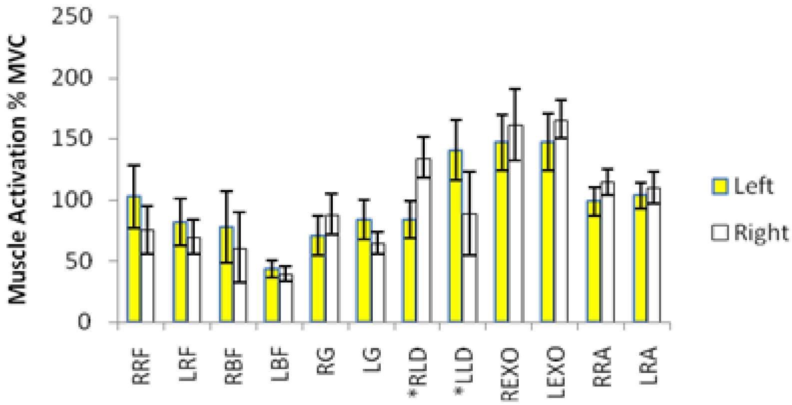

Figure 1. Comparison of peak muscle activation during the left and right paddle strokes. * depicts significant difference between left and right stroke.

The significant difference in peak activation of the latissimus dorsi during left and right strokes was characterised by higher activation in the ipsilateral muscle. This indicates the clear importance of the ipsilateral latissimus dorsi within the stroke, reinforcing the findings of Logan and Holt (1985). Correlation analysis did not, however, support this finding, indicating that the muscles of the lower abdomen were more important in the production of force. The contralateral rectus abdominus displayed significant predictive relationships with mean force during both strokes and peak force during the right stroke. Thus indicating an isometric role within the lower trunk as a stable platform against which propulsive force is developed, in what is a largely instable competitive environment.

Table 2. Correlation Results between muscular activations and force production during the left and right paddle strokes.

Muscle	Left Stroke		Right Stroke	
	Mean Force	Peak Force	Mean Force	Peak Force
Right Rectus Femoris	0.570	0.516	0.228	0.352
Left Rectus Femoris	0.323	0.431	0.719*	0.806*
Right Biceps Femoris	-0.397	-0.489	-0.384	-0.178
Left Biceps Femoris	0.042	0.390	-0.376	-0.394
Right Gastrocnemius	0.698*	0.477	0.196	0.289
Left Gastrocnemius	-0.580	-0.812*	0.270	0.197
Right Latissimus Dorsi	0.396	0.564	0.589	0.576
Left Latissimus Dorsi	0.429	0.408	-0.452	-0.346
Right External Oblique	0.249	0.143	0.439	0.442
Left External Oblique	0.801*	0.798*	0.663*	0.643*
Right Rectus Abdominus	0.855*	0.651*	0.242	0.214
Left Rectus Abdominus	0.680*	0.471	0.944*	0.955*

*** denotes significant correlation (p<0.05)**

The strong significant relationships displayed by the left external oblique with mean and peak force during left stroke (Table 2) would indicate a rotatory contribution from the muscle. Thus allowing the paddle to spend a greater duration of the stroke in contact with the water and providing greater opportunity for production of propulsive force. However, further significant relationships during the right stroke indicate a similar role to that of the rectus abdominus during the contralateral paddle stroke. This would, therefore, suggest an important role throughout the entire paddle stroke for the external oblique. However, the right external oblique displayed no significant relationships during either stroke or with either peak or mean force; although correlations were higher during the ipslateral paddle stroke (MF: 0.439, PF: 0.442, $p>0.05$). Therefore, the role of the external oblique during paddling appears to be

increasing the duration that the paddle is in contact with the water, whilst having a lesser emphasis on the production of propulsive force.
Finally, significant positive correlations were exhibited between the left rectus femoris and peak and mean force during the right paddle stroke, suggesting that the contralateral leg has an important role in force production as the hip and knee flex. Despite this relationship and the high level of ipsilateral rectus femoris activation (>80% MVC) suggesting that the lower limbs would also contribute to providing this stable base for force production, the contribution of the thigh musculature to propulsive force production during the kayak stroke has no statistical support and as such cannot be ascertained without further investigation.

CONCLUSION: Findings indicate that the musculature of the trunk and legs demonstrate clear activation throughout the paddling cycle. However, it is the muscles of the lower abdomen, namely the contralateral rectus abdominus, that have exhibited clear relationships with force output and as such are fundamental in the production of propulsive force. Furthermore, the activation of the ipsilateral latissimus dorsi has been shown to be significantly higher than the contralateral latissimus dorsi; although regression analysis displayed no significant relationship with force production. Therefore in addition to the traditional on-land resistance training, emphasis should also be placed upon recreating the instable conditions experienced on-water, to improve isometric strength of the lower abdomen in preparation for competitive performance.

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THE EFFECT OF MYOELECTRIC STIMULATION ON PERONEAL MUSCLES TO RESIST SUDDEN SIMULATED ANKLE SPRAIN MOTIONS

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This study evaluated the effect of myoelectric stimulation on peroneal muscles to resist sudden simulated ankle sprain motions. Ten male subjects performed unanticipated inversion and supination spraining motions simulated by a mechanical sprain simulator. Myoelectric stimulations with different delay time were delivered to the peroneal muscles to initiate involuntary muscle contraction and ankle joint pronation torque to resist the spraining motion. The motion was captured and analyzed by a motion analysis system, and was quantified by the reduction of maximum heel tilting angle and angular velocity. Results showed significant effect in all conditions with the myoelectric stimulation of any delay time within 15ms. The maximum heel tilting angle and angular velocity dropped from 18 to 9-13 degrees and from 200-250 to 140-170 degree/s respectively. The present corrective mechanism could be implemented in our current research to develop an intelligent sprain-free sport shoe attempting to prevent ankle sprain injury in sports.

KEYWORDS: Sports medicine, inversion, supination, injury prevention.

INTRODUCTION: Ankle ligamentous sprain as caused by sudden excessive ankle inversion or supination is one of the most common sport-related injuries (Fong et al, 2007). The two most commonly suggested aetiologies are the incorrect foot positioning at landing which generates sudden and excessive ankle inversion or supination torque, and the delayed reaction time of the peroneal muscles at the lateral aspect of the ankle to accommodate by resistive eversion or pronation torque (Fong et al, 2009a). Our research team is developing an intelligent sprain-free sport shoe which firstly senses the ankle joint motion, then identifies if a hazardous ankle spraining motion is occurring, and finally actuates a corrective mechanism to prevent an ankle sprain injury (Chan, 2006).

One of the proposed corrective mechanisms for the aforementioned purpose is to deliver a myoelectric stimulation to the peroneal muscles to trigger quick contraction. The rationale is that it could generate peroneal muscle contraction and the subsequent ankle joint pronation torque within 21-25ms (Ginz et al, 2004). This could take over the role of the slower peroneal muscles which react within 60-90ms to resist the sudden ankle torque happening within 40-50ms after the start of ankle joint twisting (Fong et al, 2009b). This study evaluated its effect by quantifying the reduction of maximum heel tilting angle and angular velocity during sudden inversion and supination spraining motions simulated by a mechanical sprain simulator (Chan et al, 2008).


Figure 1. The simulated ankle sprain test.

METHOD: Ten male subjects (age = 22.6 ± 2.4 year, height = 1.72 ± 0.04 m, body mass = 68.1 ± 8.0 kg) with clinically examined healthy ankles were recruited from the university athletic team. The

university ethics committee approved the study. Each subject performed five trials of simulated inversion test and supination test on a pair of mechanical sprain simulators (Figure 1) (Chan et al, 2008). For the simulated inversion test, the axis of the falling platform was aligned parallel to the perpendicular axis of the foot for pure ankle inversion motion. For the simulated supination test, the axis was tilted for 23 degrees medially from the perpendicular axis of the foot, which allows a natural ankle supination along the oblique axis of rotation of the subtalar joint (Hertel, 2002). In each trial, the subject was instructed to stand normally and relax with his weight evenly distributed on both platforms. Without prior notice to the subject, one of the two platforms fell freely and suddenly for 30 degrees. The fall was started by an electrical trigger operated by a research assistant when the subject was standing in a relax way as instructed. The falling sequence was randomized, and the procedure was repeated until five trials of with the right platform falling were performed.
A battery-powered myoelectric stimulation device (Figure 2a) was fabricated by the university electronics services unit by modifying a previous design (Thorsen and Ferrarin, 2009), in order to generate an adjustable electric stimulus by varying the magnitude, the duration, and the time delay of the stimulus from the start of the electrical trigger. For each subject, a pair of electrodes (Panasonic EW4312P, Japan) was attached to the muscle belly of the peroneal muscles at the lateral shank. The muscle belly was identified when the subject was instructed to perform voluntary ankle joint pronation. The skin surface was shaved and cleaned before the attachment of the electrodes. The subject was then requested to sit down and relax with both feet on the floor. A myoelectric signal of 130V was then delivered to the peroneal muscles of the subject to check if the system was well equipped to the subject, as indicated by an involuntary ankle pronation motion right after the delivery of myoelectric signal (Figure 2b). The procedure was successful for all subjects in this study. The delay time was set at 0, 5, 10 and 15ms in order to determine the maximum delay between the moment an ankle sprain starts to occur until the latest time which the device could still prevent an ankle sprain injury. Since the electromechanical delay was reported to be 21-25ms, a delay time greater than 15ms was not investigated as it could hardly catch up with a vigorous ankle sprain motion happening within 40-50ms. The activation time was set to 500ms, which is enough to cover the duration of an ankle sprain motion.

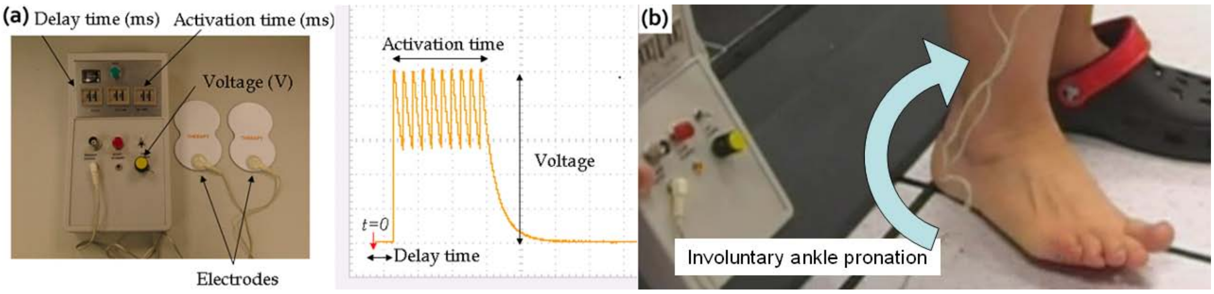

Figure 2. The myoelectric stimulation device to deliver myoelectric stimulus to peroneal muscles.

Twelve reflective markers (5mm diameter) were attached to lateral fibula head, tibial tuberosity, lateral proximal shank, medial proximal shank, anterior distal shank, lateral distal shank, medial distal shank, posterior heel, lateral heel, medial heel, medial foot and dorsal foot. Marker coordinates were recorded by an optical motion analysis system (VICON, UK) at 500Hz. The marker coordinates were filtered by Generalized Cross-Validation package of Woltring with 15Hz cut-off frequency (Woltring, 1986). A static calibration trial with the subject standing on the platforms in the anatomical position served as the offset position to determine the segment embedded axes of the shank and foot segment. The foot and shank segment were embedded with the Laboratory Coordinate System (LCS). A singular value decomposition method was employed to calculate the transformation from triad reference frame to anatomical shank and foot reference frame (Soderkvist, 1993). Joint kinematics was deduced by the Joint Coordinate System (JCS) method (Grood, 1983). Heel tilting angle was defined as the angle between the LCS vertical axis and foot sagittal plane directional axis,

and the heel tilting velocity was its change with respect to time. The maximum measurements of these two parameters were investigated. The data analysis was processed by a customized Matlab program.
Shapiro-Wilk test was conducted to check the normality of each parameter in each condition. If normality was achieved, multivariate analysis of variance (MANOVA) with repeated measures and post-hoc Tukey t-test was conducted to investigate the measured parameter in a condition statistically differ from that of control, with no myoelectric stimulation being delivered during the simulated spraining test. If normality was not achieved, Kruskal-Wallis analysis of variance and post-hoc Mann-Whitney U test was conducted instead. Statistical significance was set at $p<0.05$.

RESULTS: Figure 3 shows the maximum heel tilting angle and angular velocity in the simulated inversion and supination test, with the myoelectric stimulus delivered after different delay time. In the control conditions, no myoelectric stimulus was delivered. In both simulated inversion and supination tests, the maximum heel tilting angle dropped from around 18 degrees to 9-13 degrees, and the maximum heel tilting angular velocity dropped from 200-250 degree/s to 140-170 degree/s. Shapiro-Wilk test showed normality of all parameters, and the subsequent MANOVA with repeated measures and post-hoc Tukey t-test showed that the drop of the two parameters were statistically significant in all conditions with myoelectric stimulus with any delay time within 15ms.

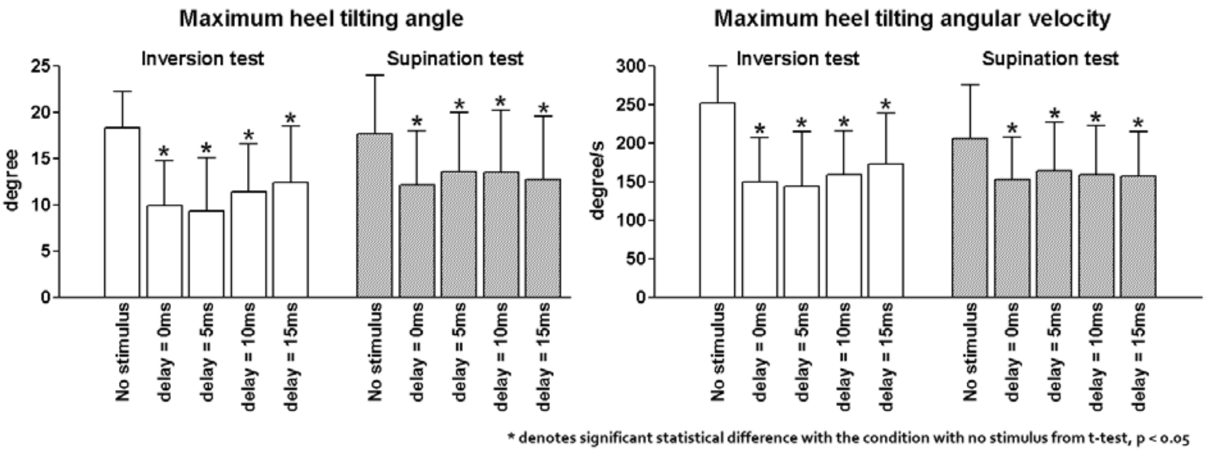

Figure 3. Maximum heel tilting angle and angular velocity in inversion test and supination test.

DISCUSSION: In this study, the myoelectric stimulation on peroneal muscles was found to be effective in reducing the maximum heel tilting angle and angular velocity in the simulated ankle sprain tests. The study was delimited to simulated but not real ankle sprain injuries, since it would be unethical and not reproducible to conduct injury trials in a laboratory. We postulated that such a sub-injury motion, which is a motion leading up to an irreversible ankle inversion or supination sprain injury, is undesirable. Therefore, it is reasonable to start the protection when such a sub-injury motion has occurred, or else an ankle sprain injury would be inevitable.
The condition with the myoelectric signal delivered after the maximum delay, which was 15ms after the start of the simulated spraining motion, was still found to be effective. As our research team is currently developing an ankle sprain identification method utilizing motion sensors to detect any hazardous ankle spraining motion (Chan et al, in press; Chu et al, in press), the results in this study suggested that we may have a maximum time period of 15ms for the sensors to successfully detect a sprain motion, and to actuate the corrective system presented in this study. Future studies should investigate the effect of the real-time sensing and identification method to serve as the trigger to activate the ankle sprain corrective mechanism, and its effect to resist the sudden ankle inversion in time.

CONCLUSION: This study showed a good feasibility of delivering myoelectric stimulation on peroneal muscles with 15ms to resist sudden simulated ankle sprain motions. This corrective mechanism could be implemented in the intelligent sprain-free sport shoe to prevent ankle sprain injury in sports.

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New Investigator Award

PROJECTED LIGHT SYSTEM FOR TRUNK SURFACE RECONSTRUCTION AND VOLUME MEASUREMENT DURING RESPIRATION

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KEYWORDS: Biomechanics, Power Crust, videogrammetry, volume, surface reconstruction.

INTRODUCTION: There is an increasing interest on developing non invasive and accurate methods to obtain torso shape and deformation during movement. Methods like inductance pletismography (Warren et al. 1989), magnetommetry (Verschakelen & Demedts 1995) and kinematical analysis (Ferrigno et al. 1994) have been proposed to access the pulmonary function based on trunk motion analysis. Measurements of body shape and dimensions are widely used on ergonomic and anthropometry designs fields (Allen et al. 2004) and to estimate body segment parameters for the analysis of human movement (Wicke et al. 2009). The aim of this work was to present a video-based method for trunk volumes measurement during the respiration by means of projected light and surface reconstruction.

METHOD: Our projected light system used four digital video cameras (JVC 9500). One pair of cameras registered the anterior trunk surface and one pair registered the posterior trunk surface (acquisition frequency of 30Hz). For each pair of cameras one multimedia projector was used to project a dense grid of circular markers on the body surface. The video stream was segmented by pre-processing techniques, morphological operators and detection algorithms to obtain 2d coordinates of the projected markers. The RGB image was converted to grayscale (8 bits) and by using thresholds the gray scale image was converted to binary format (1 bit). In the binary image, outer boundaries of objects were tracing that nonzero pixels belonged to an object and 0 pixels constituted the background. The 2D coordinates were obtained by the mean value of the outer boundary markers. The correspondence between markers on the different image projections was established by a labeling process. This process assigned the same identification (number and order) to a marker on the different image projections taking account it position and orientation in the image relative to four knowing markers. The 3d coordinates of the labeled markers were obtained by using the camera calibration parameters by means of a Direct Linear Transformation method (Abdel-Aziz & Karara 1971). After 3D reconstruction, the anterior and posterior trunk surfaces were represented by two unorganized clouds of points. The surface reconstruction tool Power Crust (Amenta et al. 2001) was used to order the cloud of points and enclose the trunk volume. The volume of the enclosed trunk was obtained from an algorithm based on the discrete form of the divergence theorem (DTA) (Alyassin et al. 1994). The surface reconstruction and volume calculation procedures were accomplished in the VTK (Visualization Tool Kit: www.vtk.org). The accuracy and the reproducibility were obtained by the comparison between plastic trunk model volume obtained by water displacement and the volume obtained by our projected light method. The volume measurement of the plastic trunk by our method was performed five times. The accuracy was defined as follows: $a^2=b^2+p^2$. Where p^2 was the variance of the experimental data and b was the bias given by the difference between the mean value (experimental measurement) and the real value (direct measurement). The method was also tested in one male subject. The participant was sitting down with the arms on a waist and he was encouraged to execute three maximal consecutive inhale and exhale cycles.

RESULTS: To demonstrate the performance of our projected light system, some reconstructed surfaces of the plastic trunk are show in Figure 1. The accuracy of the optical volume measured relative to the real value was 2.9% (SD 0.08 litters). This method also demonstrated capability for measure the trunk volume variation during breathing once was obtained a coherent signal with the respiratory cycle phases.

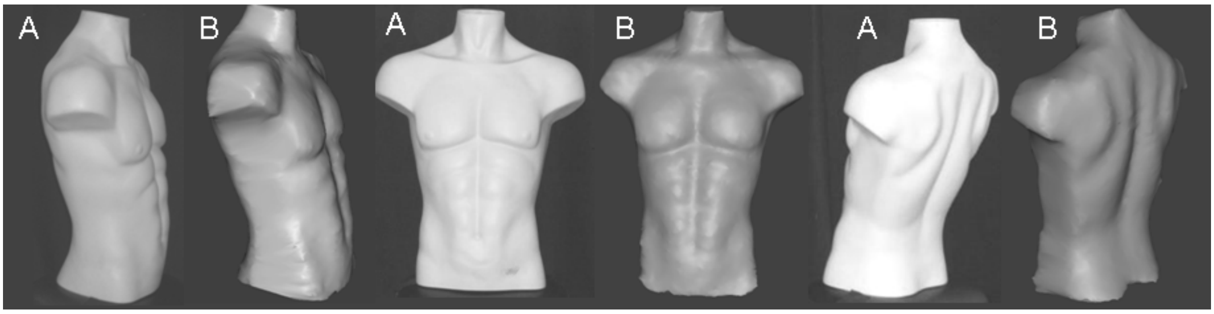

Figure 1. Examples of the plastic trunk reconstructed by our optical method (B) in comparison with the original surface (A)

DISCUSSION: Our method was capable to provide a trustworthy reconstructed trunk surface and measured trunk volume accurately and with high reproducibility. Better accuracy results for the volume measurement was only found in the literature by means of laser scan systems in static conditions (Wang et al. 2006). Another contribution of our proposed method is the segmentation and the labeling of a large number of projected markers in motion.

CONCLUSION: This novel development provides an important non-contact tool for the biomechanics field. It has potential application for the determination of body volumes to estimate body segment parameters.

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TRANSMISSION OF VIBRATION ABOUT THE KNEE

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The purpose of this study was to examine and describe effects of knee flexion angle, stance width and vibration platform frequency on the transmission of vertical acceleration about the knee. Fifteen adults were exposed to various vibration conditions while standing on a side-to-side vibration platform. Vertical acceleration data, expressed as transmission, were shown to be attenuated for all vibration conditions. A larger degree of knee flexion however, was conducive to greater attenuation about the knee. Such information may be used to develop vibration training programs with a more thorough understanding of effects of vibration.

KEYWORDS: transmission, vibration, root mean square acceleration

INTRODUCTION: The mechanistic understanding of improved performance following vibration is uncertain. Specifically, the transmission of vibration throughout the body is not thoroughly understood. When someone stands on a vibration platform, a transmission value can be calculated to assist with the description of the magnitude of the vibration imposed on body landmarks. Generally, a transmission value does not discriminate between wobbling masses and ridged bodies, yet the value may be useful in identifying the rate of vibration absorption inter- and intra-individually.
Previously, transmission has been calculated by placing accelerometers on a vibration platform and the knee (Harazin & Grzesik, 1998), head (Abercromby et al., 2007), shank and thigh (Cook et al., 2010), pelvis and spine (Mansfield & Griffin, 2002) or directly into bone (Nsiah et al., 2006). Some methods are more invasive than others, yet an acknowledged error is created when accelerometers are mounted to skin. A data correction method to eliminate effects of local tissue-accelerometer resonance from surface measurements of vibration about the spine was proposed (Kitazaki & Griffin, 1995), yet since then, correction methods have not always been used.
Recently, skin-mounted accelerometers were shown to minimally affect impact accelerations during gait compared with bone-mounted accelerometers (Nsiah et al., 2006). As such, skin-mounted accelerometers were thought of as a good predictor of skeletal impact accelerations. Another study used a three-dimensional motion analysis system to measure transmission of vibration in order to eliminate error associated with skin-mounted accelerometers (Smith, Bressel & Snyder, 2009). For field work however, laboratory procedures are often unsuitable. Therefore, field studies generally acknowledge the limitation of skin-mounted accelerometers but implement their use because of ease of operation and transportability.
Our previous pilot study had validated accelerometers for measuring vibration platform frequency (Joseph & Furness, 2009). To further that work, the purpose of this study was to explore and describe effects of knee angle, stance width and platform frequency on transmission of vibration measured by skin mounted accelerometers.

METHODS: Fifteen healthy females and males (mean age = 19.6 years ± 1.4, stature = 1.76 m ± 0.08, mass = 70.5 kg ± 10.6) freely consented to participate in the study. Participants were free from muscular injury in the previous month and had no known joint injuries. Each participant randomly received 6 bouts of side-to-side vibration delivered by a sinusoidal oscillating vibration platform (Vibro-Trainer Semi-Commercial, Amazing Super Health, AUS). Each bout lasted 60 seconds and consisted of a predetermined stance posture and stance width while vibration platform frequency was randomly assigned. Stance posture consisted of

20º, 40º and 60º knee flexion, where 0º knee flexion corresponded with full knee extension. Stance width was 10 cm and 20 cm from the axis of rotation for each leg. Vibration platform frequency was 20 Hz, 25 Hz and 30 Hz. Data were collected for five seconds at each frequency after platform steady state had been achieved. The project had University Human Research Ethics Committee approval.
The independent variables were; (1) vibration platform frequency (20 Hz, 25 Hz, 30 Hz), (2) stance width (10 cm, 20 cm) and knee flexion angle (20º, 40º, 60º). The dependent variable, transmission, was calculated from the ratio of root mean square (RMS) knee acceleration (K_{RMS}) to RMS platform (P_{RMS}). A transmission value of 1.00 represented parity between the platform and knee. A transmission value less than 1.00 represented a larger P_{RMS} than K_{RMS} (figure 1).

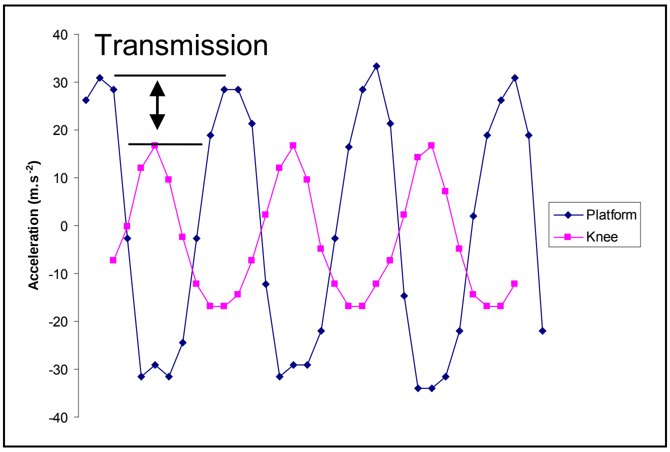

Figure 1. An example of transmission, where the maximum acceleration about the knee is less than the maximum acceleration of the platform. For this example, the transmission value would be < 1.00. Note, data are $m.s^{-2}$ rather than RMS since negative values are squared when calculating RMS values.

Two 25 *g* tri-axial accelerometers (CXL25GP3, Crossbow Technology, San Jose, USA) sampling at 250 Hz were used to quantify vibration platform vertical acceleration and knee vertical acceleration. The mass of each accelerometer was 46 gm. One accelerometer was attached to the vibration platform with double sided adhesive tape. Another accelerometer was firmly taped to the left patella of a participant to reduce skin movement. Knee flexion angle was constant for each vibration bout and checked with a manual goniometer. The accelerometer was checked for vertical alignment during each vibration bout.
Knee flexion angle was filmed with a digital camcorder (NV-GS11, Matsushita Electric Industrial Co., Osaka JPN) and digitised with Siliconcoach Pro 7 (Siliconcoach, San Francisco, USA). Small reflective markers (1 mm diameter) were adhered on the skin to the right; greater trochanter, lateral condyle and lateral malleolus. Data were filmed for each knee flexion angle independent of stance width and vibration platform frequency.
Left leg length was measured with the 'total true shortening' method (McRae, 1999). A tape was placed from the left anterior superior iliac spine to the left medial malleolus. The participant lay in the supine position. The average of two reading was recorded. These data were recorded since it was thought leg length may affect vibration transmission.
Data were imported to SPSS 17.0 for Windows (SPSS Inc., Chicago, USA). Descriptive statistics were calculated to quantify sample statistics, root mean square, knee angle and vibration transmission.

RESULTS: Sample descriptive statistics are shown in table 1. Acceleration about the knee and platform are shown in table 2. Acceleration was 28.55 $m.s^{-2}$ for 40º knee flexion, 20 cm stance width and 30 Hz platform frequency.

Table 1. Sample Descriptors

Sex	n	Stature (m)		Mass (kg)		Left leg length (cm)	
		mean	SD	mean	SD	mean	SD
Female	7	1.72	0.08	63.76	7.16	88.71	4.69
Male	8	1.80	0.07	76.35	9.71	93.19	3.93

Table 2. Knee and Platform Acceleration for Vibration Platform Frequencies, Stance Widths and Stance Postures

	20 Hz		25 Hz		30 Hz	
	10 cm	20 cm	10 cm	20 cm	10 cm	20 cm
	mean (SD)	mean (SD)	mean (SD)	mean (SD)	mean (SD)	Mean (SD)
20º	7.85 (3.63)	10.30 (2.06)	14.42 (5.10)	20.01 (2.16)	17.27 (4.91)	28.35 (7.26)
40º	8.53 (2.35)	12.07 (2.84)	10.10 (2.16)	22.96 (6.38)	24.62 (6.08)	28.55 (7.85)
60º	5.89 (1.96)	9.52 (3.14)	13.34 (5.59)	16.09 (5.40)	17.36 (3.43)	24.03 (5.49)
Platform	19.13 (1.57)		30.02 (3.92)		47.19 (4.02)	

Transmission of vibration about the knee is shown in table 3. At 30 Hz, for example, a 20 cm stance width and 60º knee flexion angle caused a transmission value of 0.53. Generally, the 60º knee flexion stance posture elicited greater transmission about the knee.

Table 4 shows variability of pre-determined stance postures during each vibration bout. Although the 20º knee flexion stance posture was most accurately maintained, variability across all stance postures was almost identical.

Table 3. Transmission of Vibration About the Knee for Various Vibration Platform Frequencies, Stance Widths and Stance Postures

	20 Hz		25 Hz		30 Hz	
	10 cm	20 cm	10 cm	20 cm	10 cm	20 cm
20º	0.73	0.67	0.87	0.74	0.57	0.68
40º	0.80	0.69	0.54	0.80	0.83	0.64
60º	0.57	0.57	0.77	0.55	0.57	0.53

Table 4. Pre-determined Stance Postures and Actual Knee Flexion Angles by 'Siliconcoach'

Pre-determined knee flexion angle	Exact	
	Mean	SD
20º	22.1º	4.2º
40º	36.6º	4.1º
60º	54.7º	4.2º

DISCUSSION: The major finding of this study was that knee angle, stance width and vibration platform frequencies concurrently affected transmission when standing upon a vibration platform (table 3). The 60º knee flexion stance posture generally caused the lowest transmission values suggesting that vibration was attenuated about the lower limbs to a greater extent than other stance postures. Knee angle however, was varied (table 4). The finding may be supported by previous work of electromyogram (EMG) data. Several authors have reported increased EMG activity of vastus lateralis during vibration with an isometric squat of 55º knee flexion (Roelants et al., 2006; Cardinale & Lim, 2003). Future research should concurrently investigate transmission and EMG activity to further understand biological responses to vibration.

Methods of this study allowed quantification of transmission despite the known limitation of skin-mounted accelerometers. The accelerometers as such, were sufficiently sensitive enough to detect such transmission attenuation. Though we did not quantify the error of

potential skin-accelerometer resonance, the error appeared consistent rather that sporadic and may explain the variance of acceleration (table 2).
Since a common goal of vibration exercise was to improve muscular strength and power in various target populations, future research should first justify a training protocol and second monitor the effectiveness of the protocol throughout the intervention. To date, reports of vibration transmission pre-, during- and post-intervention are not reported in the literature. Such knowledge may enhance understanding of biological mechanisms manipulated by vibration training.
Since we have shown that various independent variables concurrently affect vibration transmission, we propose it should be reported and monitored in all vibration protocols. Not surprisingly, effects of vibration training on transmission are unknown about the knee and other body landmarks.

CONCLUSION: Collectively knee flexion angle, stance width and vibration platform frequency affect transmission of vibration. Future research should measure vibration transmission about various body landmarks and in both healthy and sub-optimal health populations in an effort to determine the most appropriate vibration training protocol.

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ULTRASONIC MONITORING FOR THE EVALUATION OF CONDITIONING BY TRAINING SESSION FOR ATHLETES

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Non-intrusive ultrasonic detection scheme has been implemented to monitor and quantify the loading effect of training sessions on athletes. The detection is obtained along a line between two acoustic transducers with similar size and shape as stick-on electrodes. All the data is derived from the transmission time-of-flight of the ultrasonic chirp signal passing through the muscle and the ultrasonic force sensor. Muscle dynamics and force generated due to maximum isometric contraction was synchronously detected with the aid of an arbitrary function generator and a two channels transient recorder. At least 16 performance deciding parameters of athletes are quantified. The achieved spatial and temporal resolutions are ± 0.01 mm and 0.01 ms respectively. Detected movement reaction time could be used as a potential indicator to identify false-start in athletics, swimming and other necessary fields.

KEYWORDS: Ultrasound transmission in-vivo, inter-muscular force, muscular endurance, movement reaction time, ultrasonic force sensor.

INTRODUCTION: On-field performance monitoring is of high importance in sports and similar activities. Regular monitoring of the key dynamic and metabolic functions is essential to achieve dominant athletic performance. Several ultrasonic setups for body motion control have been proposed so far (patents US 5 220 922, DE 4214523 and US 7 041 062 B2). An ultrasonic monitoring scheme employing chirp technology for high resolution and rapid monitoring of the change of muscle extension was developed (Zakir Hossain et al., 2008). The system has also been used to monitor the sonic velocity variation under voluntarily activated muscle (Zakir Hossain et al., 2009). In this scheme an arbitrary function generator produces a chirped ultrasonic wave, that passing through the observed muscle. The transit signal is observed with a synchronized transient recorder. Subsequently the time-of-flight (TOF) is determined with the implemented software. Subsequent evaluation including modeling allows the determination of parameters relevant for training like: movement reaction time, muscle endurance, muscle force and power, rate of energy expenditure and other parameters of use for the optimization of training process.

METHOD: Six healthy athletes (three boys and three girls) were selected from a training camp. Their average age was 11.00 ± 0.33 and BMI 18.00 ± 0.29. The data were recorded before and after a daylong exhaustive training session. The training load was designed and organized by the professional trainer.

Flexion of the knee joint was restricted by an ultrasonic force sensor to monitor gastrocnemious muscle force and dynamics as well, for maximum isometric contraction (figure 1, left). Action was initiated by a sound beep at zero time. Monitored athletes pull back the sonic force sensor with maximum effort and hold as long as possible. The action ended with sudden withdrawal of the maximum pulling force. The position of the foot and joint angle was ensured unchanged with a suitable arrangement.
Evaluation of the collected data was performed in real time by dedicated LabVIEW software. Inflection points and conventional fitting are employed to quantify the performance variables of the monitored gastrocnemius muscle. The time from the onset of the audio stimulation to the onset of the muscle movement is interpreted as the athlete movement reaction time. Figure 1, graph (right) depicted different phases observed from the monitoring. From these, parameters relating to the athlete's muscle dynamics and energy expenditure can be obtained.

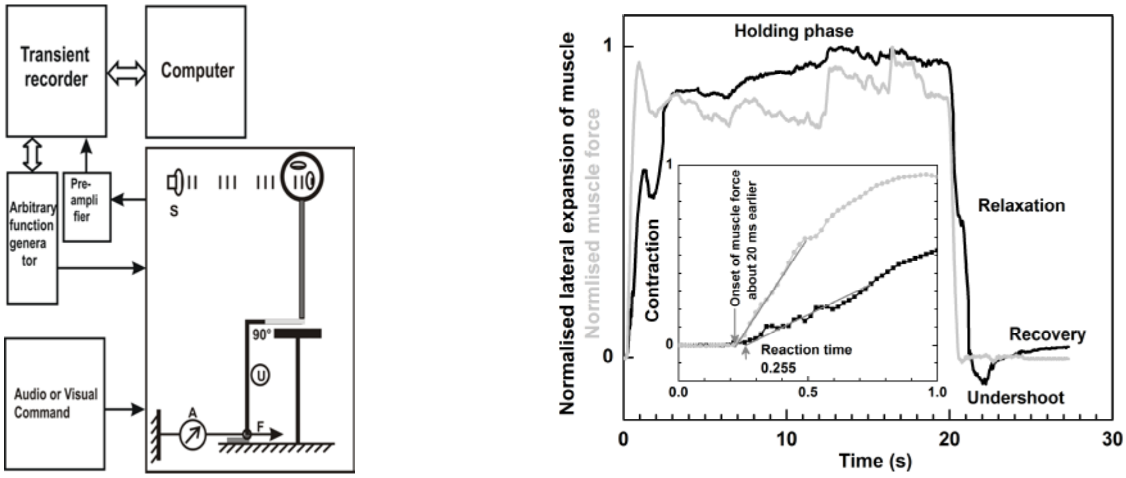


Figure 1. The schematic diagram for data acquisition setups (left) for synchronized monitoring of the muscle force and the muscle dynamics. S: audio signal, U: ultrasonic monitoring, F: delivered force, and A: ultrasonic force sensor.
Right: Graph of the obtained data. Black: muscle movement; grey: the exerted force. The inset is the analyzed transients graph demonstrating the scheme for the determination of the reaction time.

RESULTS: As illustrated in figure 1 (right) from respective readings and fits a movement reaction time of 255 ms was observed. Other observed: the maximum lateral muscle deformation 12.6 mm with variations within ± 2.1 mm, the holding phase of 18.7 s, the slope of the holding phase was about 1.1 mm s^{-1}. For the monitored muscle a contraction and relaxation speed of 3.31 mm s^{-1} and 7.76 mm s^{-1} respectively was determined. A fit to the recovery phase allowed the determination of a time constant τ = 2.27 s for recovery from the initiated action, which was a quantitative measure for the ability to recover from the post isometric tetanus effect. The muscle force onset was found approximately 20ms earlier than the muscle movement onset. The observed 17% post isometric undershoot represents an example for a scientifically relevant result related to post isometric stretch (Brenner 1990 and Alter 1996).
The figure 2 below is the graphical representation of pre and post physical loading data of an athlete. From the monitored lateral muscle dynamics and force-applied following quantitative results were obtained:

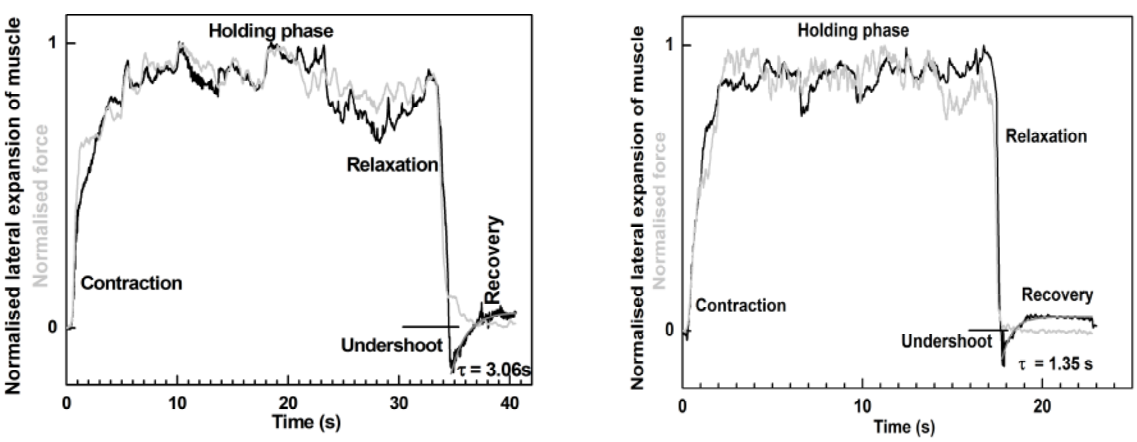


Figure 2. Displayed are the transients for muscle performance prior to (left) and after tennis training load (right) together with the applied force variation of the monitored muscle. An interpretation of the different phases is indicated.

The movement reaction time for pre- and post-physical loading were found to be 357 ms and 422 ms respectively. Contraction speed of 2.38 m s^{-1} and 2.59 m s^{-1}. A comparatively stable slope (figure 2) of 0.02 mm s^{-1} and 0.03 mm s^{-1}, were quantified from the respective holding phases. Relaxation speed 10 mm / s and 7 mm / s and the novel parameter undershoot

19.25% and 16.7% ware observed respectively for pre and post physical loading curves. Recovery time constant 3.06 s and 1.35 s were observed from the muscle dynamics curves of the monitored athlete.
Comparative results of six athletes for pre- and post-physical loading are presented below: The values are normalized to the individual initial performance to show the day long physical loading effect on the different performance deciding parameters of the monitored athletes.

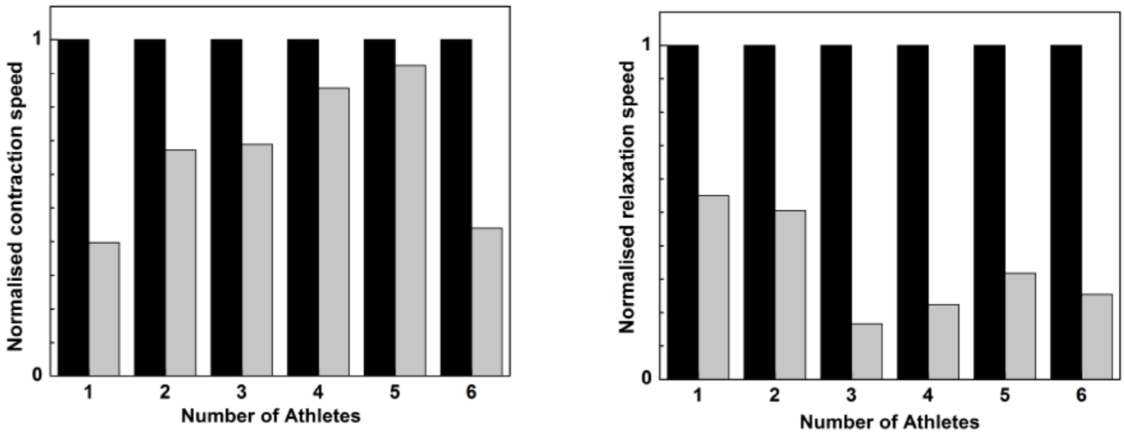
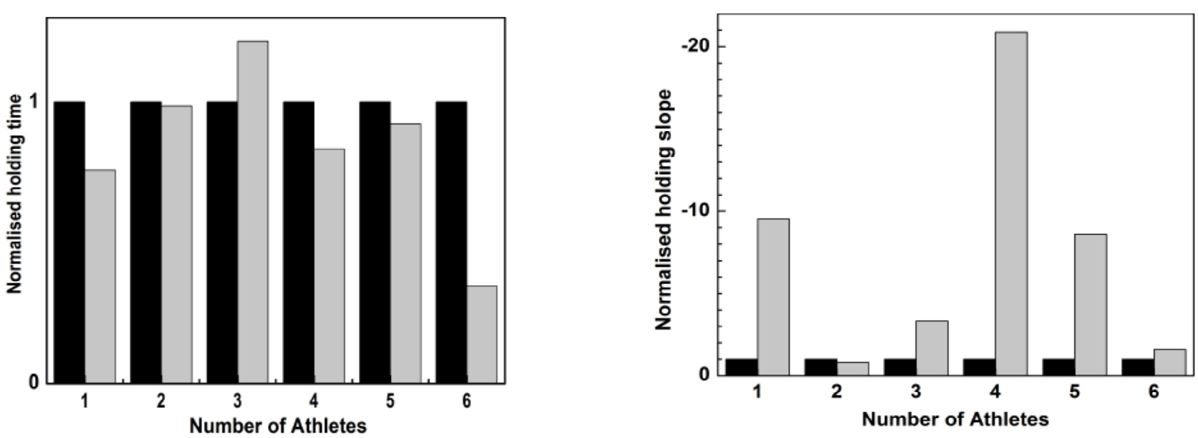


Figure 3. Displayed are the contraction (left) and relaxation (right) speed of the monitored gastrocnemius muscle for 6 different athletes. The values are normalized to the individual initial performance. The results relate to pre- (black) and post-physical loading monitoring (grey).

The bar heights of post loading muscle contraction and relaxation are comparatively lower in comparison to the individual initial i.e. pre-loading values (figure 3). That indicates muscle contraction and relaxation efficiency after a daylong training have been reduced. The holding time for athlete 6 drops substantially and athlete 3 shows an increase (figure 4, left). On the other hand the holding slopes (figure 4, right) dropped faster after the physical loading. For athletes 1, 4 and 5, holding slope droppings were substantially faster following the training



load.

Figure 4. Maximum holding time (left) and holding slope (right) for 6 different athletes prior (black) and post (grey) loading. The values are normalized to the individual initial performance.

The quantification of the holding slope for maximal isometric contraction leads to the result, that athlete 1, 4, and 5's slope falls between 10 and 21 times faster after loading. That is substantially different than observed for the other three. It is clearly indicated that the three mentioned athletes are not conditioned for endurance, since the initially observed already comparatively large holding slope dropped substantially after exercise.
The muscle force generated due to maximum isometric contraction shows deterioration after the physical loading (figure 5, left). Except for athlete 6 the muscle force shows about 20 % to 40 % deterioration.

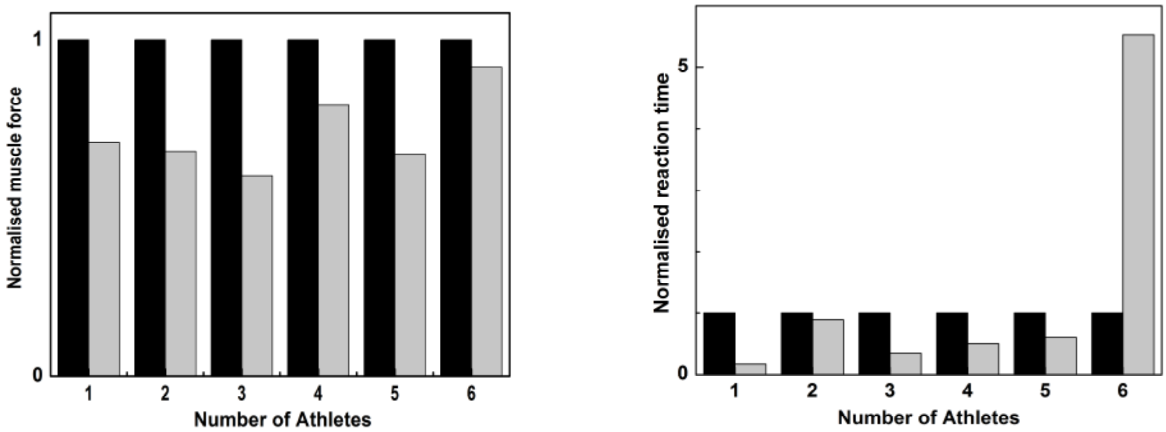


Figure 5. Maximum muscle force (left) and movement reaction time(right) for 6 different athletes prior (black) and post (grey) loading. The values are normalized to the individual initial performance.

The movement reaction time for the monitored athletes is displayed in figure 5 (right). Only athlete 6 shows a significant variation of the movement reaction time after exhaustion.

CONCLUSION AND DISCUSSION: The developed ultrasonic detection scheme is suitable for monitoring of isometric contraction which cannot be observed by high speed camera observation. The additional delay from the nerve signal to the actual movement is included which is not the case for EMG monitoring. The detection scheme is non-invasive, easily accessible and cost effective. Quantitative findings ensure its applicability in on-line monitoring of so far unobserved parameters. These include muscle contraction speed, relaxation speed, contraction impulse, relaxation impulse, slope and steadiness of the holding phase, stress, muscular endurance et cetera. The undershoot during recovery from the tetanus effect was observed for the first time in this study. Stride-length, stride-frequency, ground contact time, toe-off time, supporting phase, flying phase of running, jumping and walking are also amiable with this system. The achieved spatial and temporal resolution proves its applicability in monitoring all possible mammalian motion. This detection scheme offers easy access to assess the imparted loading effect on the athletes. Regular monitoring of this kind will help to formulate and regulate the training load to ensure maximum development of athlete's performance.

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BALANCE TRAINING ALTERS POSTURAL DYNAMICS UNIQUELY FOR STANCE ON COMPLIANT VS. NON-COMPLIANT SURFACES

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KEYWORDS: balance, proprioception

INTRODUCTON: Balance training is a common clinical modality used for improving postural control and preventing injury during sports training and participation. However, a number of empirical studies have failed to support the efficacy of balance training. One factor that may have limited the previous empirical studies is a lack of sensitivity with regard to the traditional descriptive statistics used to characterize postural control. Recent developments in non-linear dynamic analyses have led researchers to revaluate the way in which postural control is measured and understood. The advantage of nonlinear analyses for assessing postural behavior is their sensitivity to changes in the time-dependent structures of continuous postural sway. Lyapunov Exponent (*LyE*) is defined as the slope of the average logarithmic divergence of neighboring trajectories in a state space (Wolf, 1985). The purpose of this study was to evaluate the effects of balance training on postural control in a healthy population using both a traditional (position variability; as measured by standard deviation) and non-linear (Lyapunov Exponent; LyE) measure of postural sway variability.

METHOD: Twenty six healthy college-aged participants completed a six-week balance training program. The program consisted of seven balance exercise performed three times a week under the supervision of the research staff. Throughout the training period, each exercise was increased in difficulty according to the participant's willingness to proceed to more advanced levels of the exercise in concurrence with evaluation of an Athletic Trainer. This was done to mimic current physical fitness and rehabilitation training methods, as well as to ensure that each participant's postural stability remained challenged throughout the duration of the program.
Prior to and immediately following the training program, postural sway was observed by recording each participant's COP via in ground forceplate during upright bipedal stance performed on both hard and foam surfaces. Each stance trial lasted 30-sec and the COP collection rate was 100 Hz. COP *LyE* was derived using Chaos Data Analyzer (CDA) software (Sprott & Rowlands, 1992) for anterior-posterior (A-P) and medial lateral (M-L) planes independently (D=6, N=1, A=10^4). In addition, researchers also computed COP position variability, defined as the standard deviation of the entire A-P and M-L COP trajectory of each stance trial.
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RESULTS: For the M-L LyE the analysis revealed a main effect of condition (increase local stability on foam surface), $p < .000$, and no effect of phase, $p = .543$, but a condition*phase interaction, $p = .006$. Simple comparisons showed that participants exhibited increased M-L *LyE* following balance training for stance performed on the hard surface ($p = .037$), but decrease M-L *LyE* in stance performed on the foam surface ($p = .007$). For the A-P LyE the analysis showed no effect of condition, $p < .314$, and no effect of phase, $p = .484$, but a condition*phase interaction, $p = .042$. Simple comparisons showed that participants did not exhibit increased A-P *LyE* following

balance training for stance on the hard surface (*p* = .116), but did exhibit decreased A-P *LyE* in stance on the foam surface (p =.039). For the M-L COP SD the analysis revealed a main effect of condition, p < .000 (increased variability on the foam surface), no effect of phase, p = .235, and no condition*phase interaction, p = .120. Simple comparisons showed that participants exhibited no effects of training in M-L COP *SD* on the hard (p = .09) or foam (p = .415) surfaces. For the A-P COP SD The analysis revealed a main effect of condition, p < .000 (increased variability on the foam surface), no effect of phase, p = .235, and a condition*phase interaction, p = .006. Simple comparisons showed that participants exhibited no effects of training on A-P COP *SD* on the hard (p = .09) surface, but an increase on the foam (p = .002) surface.

DISCUSSION: In general, the results indicate that balance training causes a decrease in local dynamic stability (as assessed via COP *LyE*) for stance on a hard surface, but an increase in dynamic stability for stance on a foam surface. A preliminary interpretation is that inherent constraints of each surfaces requires different movement dynamics in order to promote successful upright posture that are learned during balance training. In addition, results of position variability were not generally sensitive to the effects of balance training, except for an increase in A-P position variability following training on the foam surface. Again, interpretation of this finding is that an increase in the average area of the COP trajectory might be an appropriate strategy for postural control on a compliant surface.

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QUANTITATIVE ANALYSIS OF CORE MUSCULATURE DURING TWO TYPES OF BASEBALL PITCHES: FASTBALL AND CHANGE-UP

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Previous biomechanical studies have compared kinematic and kinetics of the fastball baseball pitch to the change-up, but there is yet to be a description of muscle activations between the two pitches. With the fastball being the most common baseball pitch and the change-up being the staple off-speed pitch, it is typical for a baseball pitcher to have these two pitches in his compilation of pitches. The change-up is thrown in attempt to mimic the fastball, however has a much lower velocity than the fastball. The intention of both pitches exhibiting the same delivery is an attempt to distract the batter. Therefore it was the purpose of this study to quantitatively analyze the core musculature attached to the pelvis during both the fastball and the change-up baseball pitches.

KEYWORDS: sEMG, overhand pitching, muscle activation

INTRODUCTION: Baseball pitching is considered the most dynamic throwing task in sports. Pitching biomechanics have been investigated extensively in attempt to identify optimal pitching mechanics in terms of pitching performance (Fleisig et al., 1999; Fleisig et al., 2006; Werner et al., 2001). Based on previous quantified upper body kinetics, it has been concluded that improved muscle strength is needed in attempt to achieve adequate upper body kinetics and consequently efficient pitching performances (Fleisig et al., 1999; Sabick et al., 2004; Werner et al., 2001). It is evident that efficient transfer of energy from the lower extremity to the upper extremity is paramount in proper pitching mechanics (Stodden et al., 2001). Recently differences in kinetic and kinematic properties have been recognized in different types of pitches (Escamilla et al, 1999; Fleisig et al., 2006). However, there is limited research regarding the muscle activations that drive these kinetic and kinematic properties. Therefore, it was the purpose of our study to examine the activations of muscles supporting the lumbo-pelvic hip complex during two commonly thrown baseball pitches, the fastball and change-up.

METHODS: Twelve male Division I collegiate baseball pitchers (20.1 ± 1.5 years, 188.9 ± 4.8 cm and 87.2 ± 7.5 kg) volunteered to participate. All participants had recently finished their collegiate fall baseball season, and were deemed free of injury. Throwing arm dominance was not a factor contributing to participant selection or exclusion for this study. All testing protocols were approved by the University's Review Board.

Participants reported for testing prior to engaging in resistance training or any vigorous activity that day. Location of bilateral gluteus maximus, gluteus medius, hip adductors and external obliques were identified through palpation. Adhesive 3M Red-Dot bipolar surface electrodes (3M, St. Paul, MN) were attached over the muscle bellies and positioned parallel to muscle fibers (Basmajian and Deluca, 1985). Once all electrodes had been secured, manual muscle tests (MMT) were conducted for each muscle using techniques described by Kendall et al. (1993) in attempt to identify maximum voluntary isometric contraction (MVIC) for each muscle. Each MMT was conducted to establish baseline readings for each participant's maximum muscle activity to which all surface electromyographic (sEMG) data could be compared.

Surface electromyographic data were transmitted to The MotionMonitor™ motion capture system (Innovative Sports Training Inc, Chicago IL) via a Noraxon Myopac 1400L 8-channel amplifier. All sEMG signals were full wave rectified. Signals were smoothed based on the smoothing algorithms of root mean squared at windows of 100 ms. Throughout all testing, sEMG data were sampled at a rate 1000 Hz. In addition, all sEMG data were notch filtered at frequencies of 59.5 Hz and 60.5 Hz respectively (Blackburn & Pauda, 2009).

In addition to sEMG data, kinematic data were collected so as to event mark the phases of the pitching motion. Kinematic data were collected using The MotionMonitor™ motion capture system (Innovative Sports Training, Chicago IL). Prior to completing test trials, participants had ten electromagnetic sensors attached at the following locations: (1) the medial aspect of the torso at C7; (2) medial aspect of the pelvis at S1; (3) the distal/posterior aspect of the throwing humerus; (4) the distal/posterior aspect of the throwing forearm; (5) the distal/posterior aspect of the non-throwing humerus; (6) the distal/posterior aspect of the non-throwing forearm; (7) distal/posterior aspect of stride lower leg; (8) distal/posterior aspect of the upper stride leg; (9) distal/posterior aspect of non stride lower leg; and (10) distal/posterior aspect of non stride upper leg (Myers et al., 2005).

An unlimited time was allotted for the participants to perform their own specified pre-competition warm-up routine. After completing their warm-up and gaining familiarity with the pitching mound, data collection began. Each participant threw a series of five maximal effort fastballs and 5 changeups for strikes toward a catcher located the regulation distance from the pitching mound (18.44 m). Those data from the fastest fastball pitch passing through the strike-zone and those data from the slowest change-up pitch passing through the strike-zone were selected for analysis. Pitch velocity was determined by JUGS radar gun (OpticsPlanet, Inc., Northbrook, IL) positioned at the base of the pitching surface and directed towards home plate.

Raw data regarding sensor orientation and position were transformed to locally based coordinate systems for each of the respective body segments. Euler angle decomposition sequences were used to describe both the position and orientation of the torso relative to the global coordinate system (Wu et al., 2002; Wu et al., 2005). The use of these rotational sequences allowed the data to be described in a manner that most closely represented the clinical definitions for the movements reported (Myers et al., 2005).

Data were analyzed in the current study using the statistical analysis package SPSS 15.0 for Windows. Descriptive statistics means and standard deviations, for all sEMG were calculated for both the fastball and changeup.

RESULTS: The pitching motion was divided into five phases (stride, cocking, acceleration, deceleration, and follow through). Stride phase was described as the motion from the beginning of the pitch to stride leg foot contact (FC). The cocking phase was from stride leg FC to maximum external rotation (MER) of the throwing shoulder. Acceleration was from MER to ball release (BR). Deceleration was from BR to maximum internal rotation (MIR) of the shoulder and follow through was from MIR throughout the follow through motion. Means of %MVIC for each muscle for both pitching styles are presented in Figures 1 and 2.

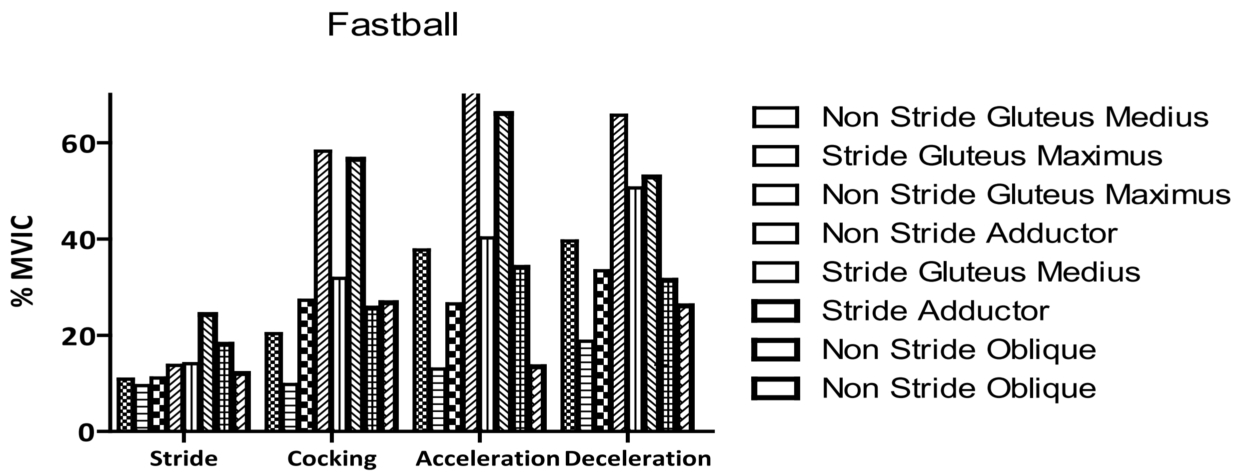

Figure 1. Muscle activations as %MVIC throughout cocking through the follow through phase while throwing the fastball.

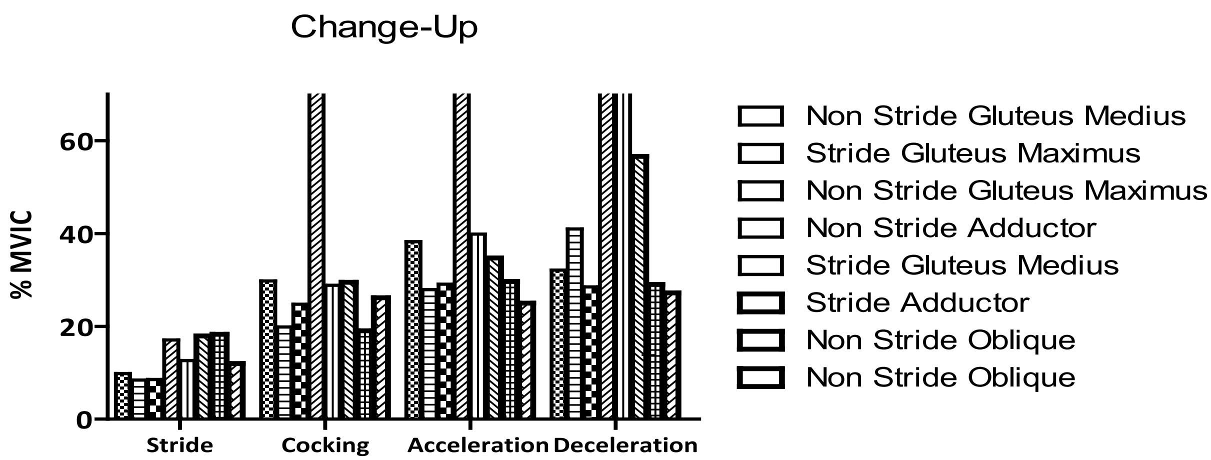

Figure 2. Muscle activations as %MVIC throughout cocking through the follow through phase while throwing the change-up.

DISCUSSION: This is the first study to investigate sEMG of the muscles supporting the lumbo-pelvic hip complex during the fastball and change-up in collegiate baseball pitchers. During the stride the hip adductors and obliques displayed a trend with the greatest activation regardless of pitch type. The fastball exhibited greater activation of the stride adductor and oblique than the non stride side during the stride phase. The activation of the gluteals, adductors and obliques during the stride is explained by their role in core stabilization while on single leg support as the pitcher is striding out into FC. As exhibited by the Trendelenberg effect, when on single leg support, the gluteus medius of the single support leg allows for pelvic stabilization to counterbalance the opposing non supported leg (Kendall et al, 1993). The cocking phase displayed the trend of greater activation of all muscles examined for both pitches as compared to the stride phase. The acceleration phase continued the trend of increases in muscle activation with the adductor muscle group generating the most activation. During the fastball pitch stride and non stride adductors were similar with the non stride side displaying greater activation. The change-up did not show adductor consistency with the stride and non stride legs having a greater difference in activation. In addition the non stride obliques demonstrated similar activations for the two pitches while the stride side oblique had decreased activation during the change-up. During deceleration the fastball and change-up deliveries demonstrated a similar trend as displayed in the acceleration phase. Follow through revealed greater muscle activation in the fastball as compared to the change-up.

CONCLUSIONS: The current study quantified and described muscle activation of the musculature controlling the lumbopelvic-hip complex while delivering two different styles of pitches. This is one of the few studies to examine the adductors and obliques during the pitching motions. We have presented only generalizations of muscle activations during both the fastball and change-up baseball pitches. It is speculated, from the data presented, that not only are the gluteals important in pelvic and torso stability (Oliver & Keeley, 2010) but also the adductors and obliques are most active in their role of pelvis and torso rotational control. Thus in attempt to target this musculature, focus should be placed training the core and torso through functional core and torso strength training protocols (Szymanski & Fredrick, 1999; Akuthota & Nadler, 2004). Additional studies are warranted in attempt to validate our results with a higher level of evidence of muscle activation and movement kinematics during the baseball pitching motion.

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ISBS 2010

Oral Session 06

Cycling

JOINT-SPECIFIC POWER PRODUCTION DURING SUBMAXIMAL AND MAXIMAL CYCLING

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KEYWORDS: Biomechanics, cycling, ergometer, muscle function, power

INTRODUCTION: Cycle ergometry is commonly used to quantify muscular work and power, and to elicit perturbations to metabolic homeostasis for a broad range of physiological investigations. Separate authors have reported that knee extension dominates power production during submaximal cycling (SUB_{cyc}; Ericson, 1988) and hip extension is the dominate action during maximal cycling (MAX_{cyc}, Martin & Brown, 2009). Changes in joint-specific powers across broad ranges of net cycling powers within one group of cyclists have not been reported. Our purpose was to determine the extent to which ankle, knee, and hip joint actions produced power across a range of net cycling powers. Based on previous reports we hypothesized that relative contributions of knee extension power would decrease and relative knee flexion and hip extension powers would increase as net cycling power increased.

METHOD: Eleven cyclists performed seated SUB_{cyc} trials (250, 400, 550, 700, and 850W) at 90rpm and MAX_{cyc} trials at 90 and 120rpm. Joint-specific powers were calculated using inverse dynamics and averaged over complete pedal revolutions and over extension and flexion phases. Portions of the cycle spent in extension (duty cycle) were determined for the whole-leg and ankle, knee, and hip actions. Relative differences in joint-specific powers across the different net cycling powers were assessed with linear regression analyses and absolute differences were assessed with paired t-tests.

RESULTS: Mean powers delivered to the right pedal were approximately one half (116±4, 200±4, 271±5, 351±5, 415±5W) of the prescribed net cycling target powers (250, 400, 550, 700, 850W, respectively) for SUB_{cyc} trials; suggesting that total power from both legs was close to the target power. Absolute ankle and hip joint-specific powers and hip-transfer power increased primarily during the extension phase whereas knee joint power increased during both the extension and flexion phases as net cycling power increased (Figure 1). Relative knee extension power decreased (r^2=0.88, p=0.01) and knee flexion power increased (r^2=0.98, p<0.001) as net cycling power increased (Figure 2). Whole-leg, knee, and hip joint duty cycle values during 250W SUB_{cyc} differed from those for MAX_{cyc} (p<0.01). Ankle joint duty cycle values during 250W SUB_{cyc} differed from those during 550, 700, 850W SUB_{cyc} and MAX_{cyc}. Absolute hip extension power increased by 19% between 90 and 120rpm MAX_{cyc} trials (356±21W vs. 423±24; p<0.01) whereas knee extension and knee flexion powers did not differ.

DISCUSSION: Our main finding was that, on average, these cyclists used relatively less knee extension and more knee flexion power as net cycling power increased. Thus, these data partially support our hypothesis and demonstrate that knee and hip joint actions used to produce power during SUB_{cyc} are relatively different than those joint actions used during MAX_{cyc}. An additional finding was that cyclists spent more time in the extension phase (increased duty cycle) during MAX_{cyc} suggesting that increased duty cycle likely serves as means to increase maximum power production. These findings support work by several previous groups that have observed duty cycle values greater than one during isolated muscle actions, animal locomotion, and single-leg cycling. Our results also suggest that hip extension power may be constrained by pedaling rate.

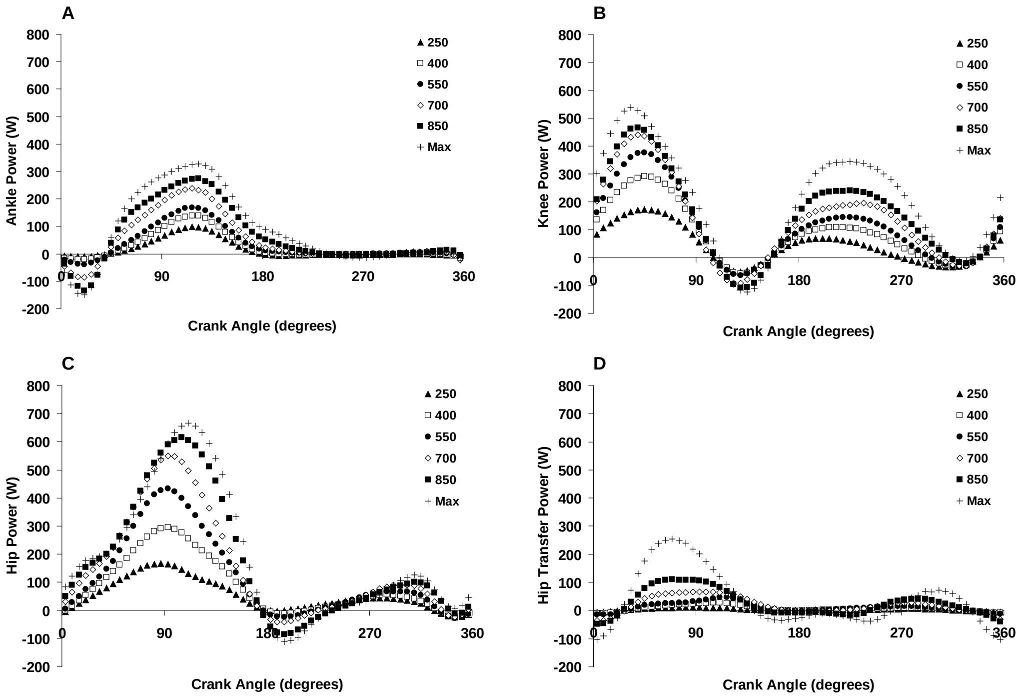

Figure 1. Power produced by ankle (A), knee (B), and hip (C) joint actions and by actions of the upper body (D) that transfer power across the hip with increasing net cycling power.

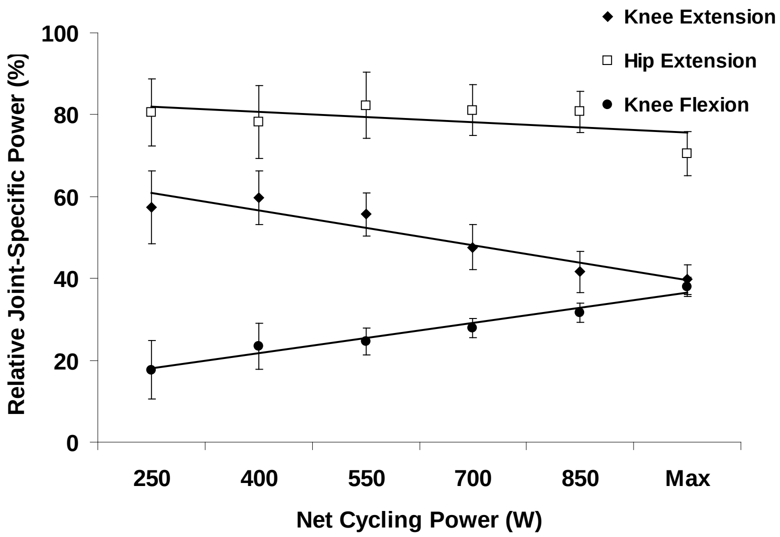

Figure 2. Alterations in relative joint-specific power with increasing net cycling power.

CONCLUSION: These are the first data to document joint-specific power production across such a broad range of net cycling powers and highlight distinct differences between SUB_{cyc} and MAX_{cyc}. These results may allow clinicians, applied sport scientists, and researchers to take even greater advantage of cycling as a laboratory tool and research model. Supported by EPSRC's Doctoral Training Grant scheme.

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THE INFLUENCE OF WORK RATE AND CADENCE ON MOVEMENT COORDINATION IN CYCLING

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This study investigated the effect of cycling cadence and work rate on coupling motion in trained male cyclists. Subjects undertook 9 pedalling bouts at various work rates and cadences (120, 210, 300 W at 60, 90, 120 rpm) and intra-limb joint coupling motions were examined using a continuous relative phase (CRP) analysis. The hip/knee (HK) coupling motion was significantly more in-phase during the 90 and 120 RPM trials compared with the 60 RPM trial (recovery phase). Similarly the knee/ankle (KA) coupling motion was significantly more in-phase in the 120 RPM trials than the 60 or 90 RPM trials (propulsive phase). No differences were found between work rate conditions. The results suggest for higher cadences the resulting movement patterns are more stable and consequently more economical. Cyclists should therefore seek to maintain a higher cadence.

KEYWORDS: cycling, coordination, cadence, work rate, economy

INTRODUCTION: In a kinematic chain the motion of one segment subsequently influences the motion of an adjacent segment, and therefore the study of isolated joints does not effectively capture the complexity of the coordinated motion of components of the body (Bartlett et al., 2007). The consideration of the coupling relationship between segments may therefore be crucial in the analysis of human movement.

There is conflict within the cycling literature regarding the most economical cadence, defined in this study as that which is associated with the lowest metabolic cost at a given work rate. This is due in part to its work rate-dependent nature (Ansley & Cangley, 2009). Li (2004) found as cadence increases there is an added influence of the inertial properties of the limbs, which consequently affects neuromuscular coordination. Changes in the coordination patterns utilised by cyclists as a result of changes to the work rate and /or cadence may therefore have an effect on their economy.

A key component in the analysis of movement coordination is the role of variability within the system under investigation (Wilson et al., 2008). Movement variability is important in skills where the adaptability of complex motor patterns is necessary within dynamic performance environments (Button et al., 2006), enabling athletes to adjust to both intrinsic and extrinsic factors (Bradshaw & Aisbett, 2006). However, in skills where tight task constraints are imposed, such as in cycling, there is likely to be a reduced requirement for flexibility and any variability present in the system may therefore be indicative of an inconsistent performance. In support of this Chapman et al. (2009) concluded that elite cyclists had greater consistency of inter-joint coordination compared with novice cyclists.

The aim of this study was therefore to investigate the affect of the work rate and cadence on the coordination exhibited by trained male cyclists and the subsequent implications for training and competition in terms of adopting the most economical strategy.

METHODS: Six trained male cyclists were recruited for the study. All subjects gave written informed consent and were free from injury at the time of the study. Using a two-scanner Cartesian Optoelectronic Dynamic Anthropometer (CODA) motion analysis system three-dimensional kinematic data were collected at a sampling rate of 100 Hz. Exercise was performed on a Monark braked cycloergometer. Twenty-three active markers of 2-mm diameter were attached to the right lower limb and the pelvis. The markers were located on the following anatomical landmarks: 5th metatarsal head, 1st metatarsal head, lateral malleolus, medial malleolus, heel, medial and lateral knee epicondyles, greater trochanters, anterior superior iliac spines, iliac crests and posterior superior iliac spine. The remaining markers were attached to polystyrene plates which were placed on the distal thigh and shank. Each plate contained a cluster of 4 markers. An additional marker was placed on the pedal axis in order to identify individual revolutions.

Subjects undertook 9 pedalling bouts in a randomized order at various work rates and cadences (120, 210, 300 W at 60, 90, 120 rpm). Subjects were instructed to reach the required cadence (visual feedback provided by digital RPM-meter) and maintain this for at least 10 seconds before data recording commenced. Data were recorded for a minimum of 20 s. A minimum of a one minute of recovery was given between trials. For each trial a total of 10 consecutive revolutions within $\pm$ 2 rpm of the required cadence were selected for subsequent analysis. A complete revolution was the time from top dead centre (TDC) to subsequent TDC. TDC was defined when the pedal marker reached its maximal value in the z-axis. Visual 3D motion analysis software (C-motion) was used to calculate 3D joint angles according to a method outlined by Grood and Suntay (1983). Prior to this raw coordinate data was smoothed using a fourth order Butterworth digital filter with a cut-off frequency of 8 Hz. The cut-off frequency was selected using Winter's (1990) residual analysis technique. Only sagittal plane data were used for further analysis. The time series of each joint angular position and velocity was assessed on a revolution-by-revolution basis and interpolated to 100 data points using a cubic spline technique. The intra-limb joint coupling motions were assessed for each revolution using a continuous relative phase (CRP) analysis, which was calculated using the angular position and velocity profiles of the relationship between the joint actions (Dierks and Davis, 2007). CRP was assessed for 2 intralimb couplings: ankle plantarflexion/dorsiflexion - knee flexion/extension (KA) and knee flexion/extension - hip flexion/extension (HK). The joint angle and angular velocity data were normalised to the maximum and minimum of the athlete-specific data set according to the procedure presented by Hamill et al. (1999). The CRP time histories for the sagittal plane KA and HK joint couplings were determined by quantifying the difference between the phase angle of the distal and proximal joint at each time interval. CRP describes the relationship between two oscillators in the phase-plane domain. A CRP of 0º indicates in-phase coupling, meaning the phase angles for the two motions are identical, and a potentially stable coupling pattern exists as they are behaving similarly. As the CRP increases from 0º in either a positive or negative direction, the two motions become more out-of-phase and are behaving in a less similar fashion. Individual averaged time histories for the CRP and the associated variation of CRP (CRPv) were determined across all revolutions for each trial using the mean CRP and associated standard deviation (SD) respectively at each time point. Time histories for the group averaged CRP and CRPv were determined as the average across each time point of the individual-specific CRP and within athlete CRP averaged profiles, respectively. This was repeated for each condition. For each coupling, the effects of cadence and work rate (and the subsequent interaction effects) on CRP and CRPv were determined using a 2-way repeated measures ANOVA. Where significant interaction effects were identified, post hoc analyses were employed to examine where the significant differences existed. In addition, differences in CRP and CRPv between the propulsive and recovery phases of the revolution were examined. Significant differences were accepted at $p < 0.05$.

RESULTS: No significant differences in CRP or CRPv were found between work rate conditions for either KA or HK. Significant differences in CRP were found between the propulsive and recovery phases for both couplings with a more in phase motion being displayed during the propulsive phase (propulsive *vs* recovery; KA, 27.4º ± 8.9 *vs* 48.5º ± 20.5, p = 0.000; HK, 22.5 º ± 6.7 *vs* 32.5º ± 6.8, p = 0.000). Significant differences in CRP were also found between the cadences for the HK coupling during the recovery phase with the 60 RPM trial displaying more out of phase motion than either the 90 RPM or 120 RPM trials (36.4º ± 3.5 for 60 RPM *vs* 33.3º ± 3.4 for 90 RPM, p = 0.030 and 27.9º ± 13.6 for 120 RPM, p = 0.026). Differences in CRP for the KA coupling were found during the propulsive phase only with the 120 RPM trials displaying significantly more in phase motion than either the 60 RPM or the 90 RPM trials (19.2º ± 12.3 for 120 RPM *vs* 30.0º ± 7.1 for 60 RPM, p = 0.011 and 33.1º ± 7.4 for 90 RPM, p = 0.024). There were no differences in CRPv across the cadence conditions for the HK coupling however in the KA coupling a significantly higher CRPv was displayed during the recovery phase in the 60 RPM trials compared to either the

90 RPM or 120 RPM trials (16.6º ± 7.6 for 60 RPM *vs* 11.6º ± 6.5 for 90 RPM, $p = 0.005$ and 8.9º ± 4.1 for 120 RPM, $p = 0.003$).

DISCUSSION: The intra-limb coupling motion of trained male cyclists was quantified for the propulsive and recovery phases of cycle revolutions at three different work rates (120, 210 & 300 W) and three different cadences (60, 90 & 120 RPM). The more out of phase motion of both the KA and HK couplings during the recovery phase suggests a less stable motion than in the propulsive phase as out of phase motion has previously been considered to reflect a less stable coordinative state (Scholz, 1990). When considering the effect of cadence on the CRP, a more out of phase movement pattern was displayed during the 60 RPM trial for the HK coupling (recovery phase) and a more in phase motion was displayed during the 120 RPM trial for the AK coupling (propulsive phases). Both these findings suggest the higher the cadence the more stable the resulting movement pattern. A stable coordinative pattern is able to be maintained despite perturbations to the system (Robertson, 2001) and according to Zanone et al. (2003), the more stable a movement pattern is, the lower the metabolic cost required to maintain the pattern at a given level of stability. This suggests that the coordination patterns exhibited at the higher cadences are more economical. This support for the use of a higher cadence is in agreement with Lucia et al. (2004) who found that for a fixed work rate, economy improves at increasing pedalling cadences and this improvement was attributed to a lower motor unit recruitment. The higher CRPv in the 60 RPM trial for the KA coupling during the recovery phase suggests a less consistent movement pattern and according to van Emmerick and van Wegen (2000) this is a sign of a less stable system. This is consistent with the CRP findings and also suggests that the variability present in the system is not beneficial to performance, something which has previously been suggested by Chapman et al. (2009). The fact that no differences in coupling motion were identified between work rates may be surprising given the significant differences between cadences and the interdependent relationship of work rate and cadence. However, the work rates investigated in this study were limited and greater ranges may be required to identify any differences which exist.

CONCLUSION: The results of this study suggest that changes in cadence may result in changes in stability and subsequently the economy for a given coordination pattern. This may have implications for both training and competition. Specifically the results support the use of a higher cadence. In addition, the less stable pattern identified during the recovery phases potentially highlights the need for further consideration of this phase by coaches. This study has been limited to intra-limb coordination however future work investigating inter-limb coordination is advocated.

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A COMPARISON OF PEDALING MECHANICS IN EXPERIENCED POSE AND TRADITIONAL CYCLISTS

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The purpose of this study was to describe the mechanical differences between two experienced male Pose cyclists and traditional cyclists. Pose cycling requires a specific set-up that attempts to place the centre of mass over the pedal to increase the non-muscular force/power component during the downstroke. Results revealed that for the Pose cyclists the centre of mass was closer to vertically above the pedal when in the horizontal forward position. Non-muscular contributions to pedal power tended to be greater in the Pose cyclists compared to the traditional cyclists. In addition, we found joint differences in the power contribution to pedal power in the traditional cyclists who relied more heavily on contributions from the muscles spanning the knee joint, whereas Pose cyclists had greater ankle and hip power contributions.

KEYWORDS: GRAVITY, CENTRE-OF-MASS, TECHNIQUE

INTRODUCTION: When performing sporting movements, it is important to employ movement techniques that are beneficial with respect to the goal of the task. Depending on the movement to be performed a good technique can be characterized by maximizing parameters such as metabolic efficiency or the mechanical effectiveness of the applied forces.

In cycling, the role of pedaling technique has been debated. While cyclists can consciously apply muscular forces more effectively to the pedals (Mornieux et al., 2008), increases in mechanical effectiveness are not associated with increases in metabolic efficiency (Korff et al., 2007).

Training with uncoupled cranks teaches cyclists to apply muscular forces more effectively (Williams et al., 2009). However, uncoupled cranks are not allowed in competition, hence, the need for a technique that uses coupled pedals.

Another pedaling technique that has been promoted in the popular literature is the so called Pose technique (Romanov, 2008). The Pose cycling technique aims to increase the gravitational contribution to the pedal force by placing more body weight over the pedal by way of a specific bicycle set-up. In addition, cyclists are encouraged to consciously use their body weight to enhance pedal power during the downstroke. Hence, Pose cycling aims to increase the non-muscular contribution to pedal power during the downstroke (Romanov, 2008).

Non-muscular forces can be quantified through inverse dynamics (Kautz & Hull, 1993), and it has been suggested that minimizing non-muscular forces plays an important role when selecting a preferred pedalling cadence. Thus, an aim of the present study was to describe the non-muscular power contributions for Pose cyclists and traditional cyclists during the downstroke of the crank cycle. A further aim was to describe differences in their pedalling technique by comparing the effective forces and the joint power contributions to total power.

METHODS: Two male Pose cyclists (age: 43 and 55 years, stature: 1.83 and 1.73 m, leg length 0.96 and 0.91 m, mass: 75.0 and 78 kg) and two traditional male cyclists (age: 35 and 35 years, stature: 1.85 and 1.83 m, leg length 0.98 and 0.95 m, mass: 74.5 and 82.0 kg) participated in the current study. All of the participants were experienced cyclists (>20 years) and considered to be exemplars of their respective techniques. Ethics approval for all procedures was obtained from Brunel University and all participants provided written informed consent. Steady state cycling at 200 W was performed at three cadences (70, 90, and 110 rpm) on an electromagnetically braked cycle ergometer (Velotron, Racermate,

USA), which was calibrated before the start of the study using a dynamic calibrator (Model 17801, Vacumed, CA).
Pedal-reaction forces were measured at 960 Hz using a custom-made force pedal with two triaxial piezoelectric force sensors (Kistler, model 9251AQ01). Pedal angle and crank angle were measured at 120 Hz using an eleven-camera motion-analysis system (Motion Analysis, Santa Rosa, CA). Force and kinematic data were low-pass filtered (second-order Butterworth) using cutoff frequencies of 20 and 10 Hz, respectively. Pedal angle and crank angle were calculated from the kinematic data. The force data (forces normal and tangential to the pedal) were down sampled to match the kinematic data. Using the kinematic and force data, the force components perpendicular and radial to the crank were calculated. Hip transfer power was defined as the dot product of hip reaction forces and hip linear velocity). The joint powers for the hip, knee and ankle joints were derived using standard inverse dynamics techniques (Hull and Jorge, 1985).
All joint angular positions were obtained from the motion analysis data. Linear and angular velocities and accelerations of the limb segments were determined by finite differentiation of position data with respect to time. Using these geometrically determined kinematics together with pedal forces, we determined the joint moments at the ankle, knee and hip joints. These moments were multiplied by the corresponding joint angular velocities to obtain joint powers. Hip transfer power was calculated as the dot product of hip reaction force and hip linear velocity (Broker & Gregor, 1994). The hip transfer power was included in this calculation to account for muscular power that resulted in linear motion of the hip (i.e. not angular motion of the joints of the lower limb). For all participants the derived power profiles were representative of fifty complete revolutions within the exercise bout. Pedal power was defined as the dot product of pedal force and pedal linear velocity. For each participant mean pedal power was calculated as the average pedal power over the corresponding pedal power profile. Hip, knee and ankle joint powers were averaged over the corresponding joint power profiles. Muscular pedal power was calculated by adding the muscular joint powers (ankle, knee, hip and hip transfer) at each point. Non-muscular pedal power was calculated by subtracting the sum of muscular joint powers and hip transfer power from total pedal power at each time increment. Joint powers as well as muscular and non-muscular pedal powers were normalized by mean pedal power to allow for meaningful comparisons. Each participant's bike set up was recorded (Table 1).

Table 1. Participants' cycle ergometer set ups

Bicycle setup: Participants	crank length (cm)	seat height crank-saddle (normalized)	seat angle (º)	handle bars position to saddle A-P (cm)	S-I (cm) (normalised)
Pose 1	17.5	0.39	77.5	0.34	0.01
Pose 2	17.5	0.40	80.0	0.31	0.02
Trad 1	17.5	0.45	73.0	0.39	0.07
Trad 2	17.5	0.45	74.0	0.37	0.04

A-P = anterior-posterior S-I = superior-inferior

RESULTS: The cycle ergometer set up differed between the Pose cyclists and the traditional cyclists. When normalised by body height, saddle length as well as vertical and horizontal handle bar position with respect to the middle of the saddle were smaller for the Pose cyclists compared with the traditional cyclists (table 1). In addition, the Pose cyclists used a steeper seat angle.

Table 2. Pose cyclists (*N=2*) and traditional cyclist's (*N=2*) parameters for one pedal cycle ($\bar{x}$ of fifty cycles)

	Averaged non muscular contribution during the downstroke (normalised)	Maximum non muscular contribution (normalised)	Ankle Power contribution (%)	Knee Power contribution (%)	Hip Power contribution (%)
Pose 70 rpm	1.12	2.09	7.91	48.13	37.55
Pose 90 rpm	1.07	2.16	6.35	60.32	27.83
Pose 110 rpm	1.24	2.49	8.62	62.04	23.30
Trad 70 rpm	0.78	1.51	10.16	48.21	31.73
Trad 90 rpm	0.84	1.91	7.53	63.71	19.99
Trad 110 rpm	1.06	2.35	2.61	98.59	-6.55

In agreement with our hypothesis, the non-muscular contribution for Pose cyclists was larger than that for traditional cyclists. In addition, the joint power contribution to total power differed between the groups (Table 2). The knee power contribution to total pedal power was greater for the traditional cyclists, whereas the ankle and hip contributions were smaller. This effect was more apparent at the higher cadences. This difference in coordination patterns is illustrated in Figure 1.

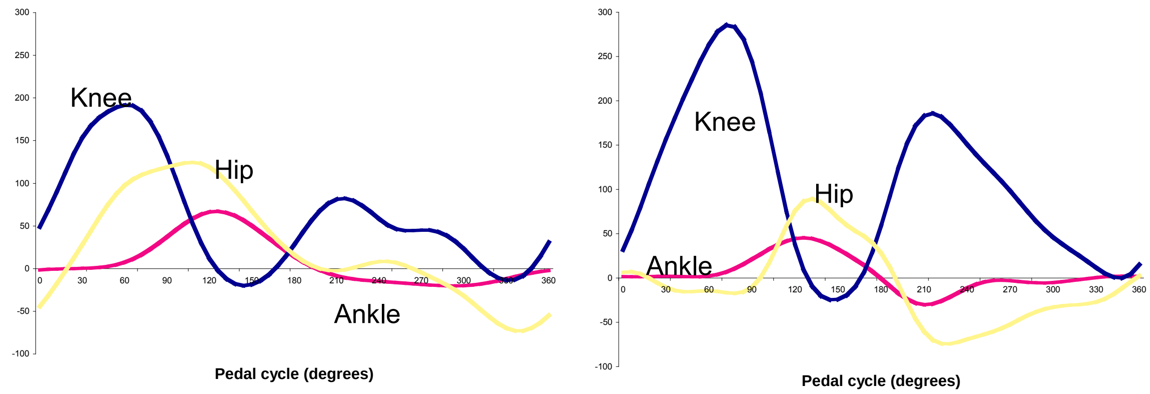

Figure 1. Pose (left *N=2*) and traditional (right *N=2*) ankle, knee and hip joint powers at 110 rpm

DISCUSSION AND CONCLUSIONS: Our findings demonstrate that the Pose cyclists produce a greater non-muscular contribution to pedal power during the downstroke when compared to traditional cyclists. This difference is accompanied by differences in the relative joint power contributions to total power. Hence, Pose cyclists may have an advantage in triathlon and distance cycling while traditional cyclists may be more effective in sprint cycling. Note that a greater non-muscular contribution during the downstroke (i.e., a greater gravitational assist) will result in a greater negative (mechanically ineffective) non-muscular contribution during the upstroke, it remains to be seen if the observed difference in technique (as described by mechanical parameters) can translate into improvements in performance and/or efficiency. This should be the subject of future research.

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FORWARD SEAT POSITION EFFECTS ON CYCLING KINEMATICS

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The aim of this study was to identify the effects of fore-aft position of the seat on kinematics during a submaixmal cycling session. Each of four recreational athletes (2 road cyclists, 2 triathletes) completed a 20-km simulated course under two different seat positions: tip of seat 5 cm in front and 5 cm behind the crank axis. Trunk and leg kinematics were determined using three-dimensional motion capture system. Bringing the seat position forward resulted in a more extended trunk-hip region (116±5° vs.122±3° of flexion); however, the source of the extension varied among individuals arising from the pelvis and the thigh in different participants. The knee joint angle range of motion and pattern were unaffected by the seat position. These results imply that participants used different muscle activation strategies in response to the change in riding position.

KEYWORDS: biomechanics, geometry, triathlon.

INTRODUCTION: Triathlon is a sport that involves three different modes of endurance events (swimming, cycling, and running) performed consecutively. Due to the nature of the sport, athletes are required to train and perform well in all three disciplines to become successful. Effective transitions between disciplines are considered one of the keys for a better performance. Although training specifically targeting the cycle-run transition has been attempted, triathletes often express that cycling impairs their running performances. Their testimonies are confirmed by literature that examines the effects of a prolonged cycle on subsequent running. Prior cycling is reported to affect running performance while the effects of swimming on cycling and running performances are considered minimal. An experiment involving a run-cycle-run session reported increased metabolic cost by 2.3 ± 4.6% during a run following a cycling bout compared to a pre-cycling run among non-elite triathletes (Millet et al., 2000). The change in cost demand is thought to be explained by both physiological and biomechanical changes. A pre-run cycling session is shown to change the kinematics (Gottschall & Palmer, 2000) and the muscle activation pattern (Heiden & Burnett, 2003) during running. One of the strategies that have been implemented by triathletes to lessen the effects of cycling on their running performance is changing the bike frame geometry. Specifically, triathletes use steep seat post angles that are more vertical (about 80°) than that of conventional road-racing bikes (between 70 to 76°). Seat post angle affects the seat's relative position to the crank axis. The more vertical seat post in a triathlon-specific bike frame places the rider more directly above the crank axis. This riding position has been shown to improve cycling performance and subsequent run performance (Garside & Doran, 2000) . This position places the cyclist in a more extended lower limb position that appears to influence force production (Ricard et al. 2006) and muscle activation patterns (Brown et al., 1996; Mestdagh, 1998; Ricard et al., 2006) during a pedal cycle. Manipulating trunk tilt alone resulted in changed muscle activation pattern in the leg, and that was likely due to modified joint and segmental kinematics (Savelberg et al., 2003). Despite its effect on performance and muscle activation pattern, the cycling kinematics with forward seat riding position has not been investigated. Therefore, the purpose of this study was to examine the effects of the forward seat position on kinematics during submaximal cycling.

METHOD: Four recreational athletes (3 male – 2 cyclists and 1 triathlete; 1 female triathlete; height: 1.76±0.09 m; mass: 69.4±12.8 kg) who regularly train on either road or triathlon bikes (71.3±35.7 miles/week; 40 mile/week minimum) volunteered and provided written informed consent to participate in the study. Both cyclists and triathletes were included so that the preferred bike geometry types were counterbalanced. In accordance with the study protocol, all subjects participated in at least one event sanctioned by USA Cycling or USA Triathlon during the past 10 months and were free of injury or illness at the time of study. All

procedures were approved by the local institutional ethics review board. The stationary bike set-up was based on five measurements of the participant's bike: seat post length (SPL: distance between the crank axis and the rail of the seat); reach length (RL: horizontal distance between the tip of the seat to the center of the handlebar); seat position (SP: horizontal distance between the tip of the seat and the crank axis); handlebar height (HBH: vertical distance from the tip of the seat to center of the handlebar); crank arm length (CAL), and; number of the teeth on all chain rings.
A stationary bike equipped with an electromagnetic resistance unit (Velotron Elite, RacerMate Inc., Seattle, WA, USA) was set-up to match each participant's personal bike measurements with two variations on the SP horizontal distance from the crank axis of 5 cm in front (road/shallow; ROAD) and 5 cm behind (triathlon/steep; TRI)(Figure 1). The Velotron 3D software allowed the cyclist to ride a virtual course using the virtual gear function that was set to gear ratio options available on the participant's own bike. A clip-less pedal-shoe interface (model X, Speedplay, San Diego, CA, USA) was used to simulate a more realistic riding condition. A pair of aerobars was used to control the riding position. Prior to the actual testing session, each participant rode the stationary bike as long as necessary to become accommodated to the bike. Forty-five retro-reflective markers were placed on the pedals, crank arms, crank axis and the cyclist's body.

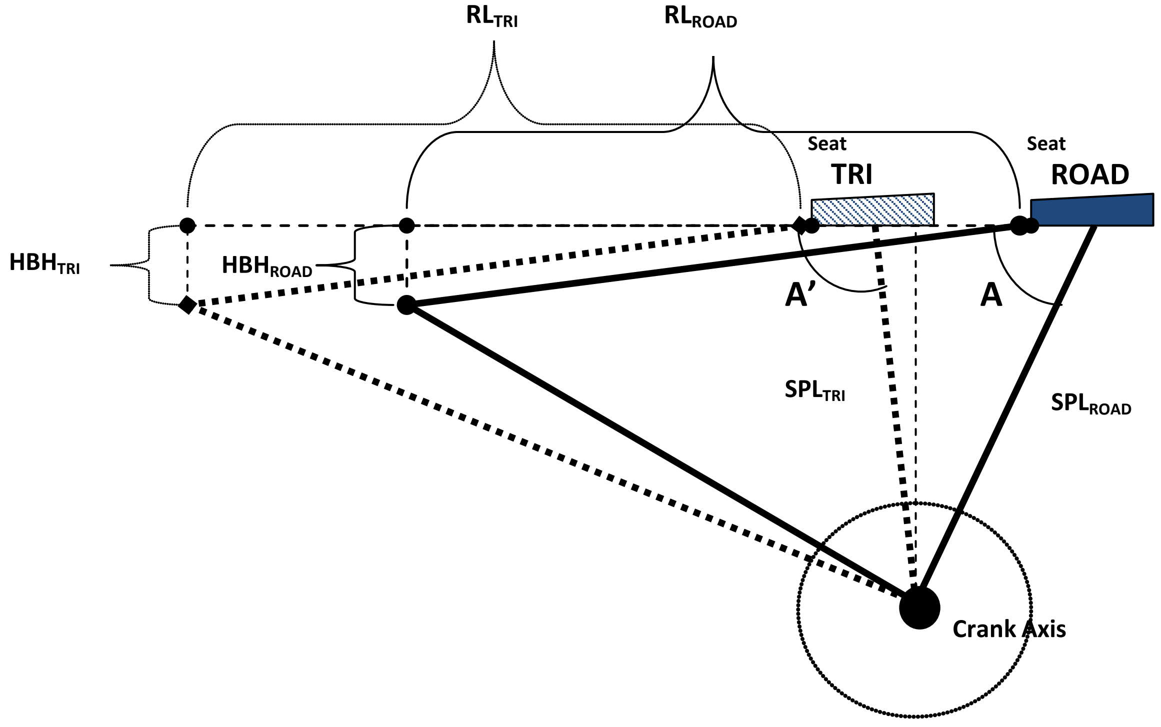

Figure 1. Geometric scheme of the seat position setup for a triathlete bike (TRI; tip of seat 5 cm in front of crank axis) and road bike (ROAD; tip of seat 5 cm behind crank axis). Locations for crank axis, handlebar height (HBH), reach length (RL), seatpost length (SPL) and seat angle (TRI is A'; ROAD is A) are illustrated.

After participants performed their standard warm-up, each rode the ROAD and TRI bike setup conditions in random order at 5-14 days apart. Each participant followed a similar exercise, dietary, and rest routine days into both testing sessions to avoid undesirable effects. The participants rode a 20-km virtual course utilizing the virtual gear to maintain their preferred cadence, with the objective of completing the course in as fast a time as possible. The subject maintained an aerodynamic position with his or her forearms resting on the elbow pads of the aerobars. At 5 km a 30-second data collection of the markers were recorded at 100 Hz using 10 Eagle cameras using Cortex software v1.0 (Motion Analysis

Corp., Santa Rosa, CA, USA). The three-dimensional coordinates of the markers were filtered using a fourth order zero-rag Butterworth filter at cut-off frequencies of 8 Hz (leg markers), 6 Hz (pelvis markers), and 4 Hz (trunk markers). Segmental (relative to a vertical for the trunk and pelvis, and relative to the horizontal for foot/pedal) and sagittal joint (relative to the proximal segment for the trunk, hip, knee and ankle) angles were calculated using Visual3D v4.0 (c-motion, Germantown, MD, USA). All full pedal cycles' for the right limb during the 30-second were interpolated to 360 data points and averaged. Data are presented as means±SD.

RESULTS AND DISCUSSION: Preserving bike set-up measurement while altering horizontal seat position primarily changes the angles at the base of the seat (A and A' in Figure 1 and in Table 1). The angle was larger for the TRI than that for the ROAD bike setup. Increases in this angle are thought to 'open the hip' (i.e. greater extension) to make the muscle length more favorable for the hip musculature (Mestdagh, 1998; Garside & Doran, 2000). The current data support this idea. All of the study participants decreased in relative hip flexion angle with the TRI seat position (Table 1). However, the decreased overall trunk-thigh flexion angle appears to be individualized with different joint angles being altered by the individuals. An experienced triathlete (sub1) accomplished this by decreasing the absolute trunk forward lean, while increasing the anterior tilt of the pelvis. On other hand, one of the participants who was a road cyclist (sub3) extended the trunk-thigh angle while maintaining both anterior pelvic tilt and forward lean of the trunk. Although both of these individuals 'opened up' the hip, one cycled with a greater pelvis-hip angle (sub1) while the other cycled with a lower pelvis-hip angle (sub3) during the TRI condition. Changing the orientation of a single segment (trunk) has shown to affect the muscle length of most of the leg muscles (Savelberg et al., 2003). Therefore, these different strategies used by the participants could possibly suggest different uses of the hip and back musculatures. The commonly accepted definition of the hip angle, the angle between the thigh and the trunk, may not fully describe what might be happening at the hip and pelvis.

Table 1. Seat angle and participant angles (hip, anterior trunk lean, and anterior pelvic tilt) for two bike set-ups (ROAD is tip of seat 5 cm behind crank axis; TRI is tip of seat 5 cm in fron of crank axis) for four participants (sub1 to 4) during a submaximal cycling session.

	Seat angle (°)		Hip angle (°)		Anterior trunk lean (°)		Anterior pelvic tilt (°)	
	ROAD	TRI	ROAD	TRI	ROAD	TRI	ROAD	TRI
Sub1 [T]	79	86	126±18	116±18	77±1	73±1	2±1	16±1
Sub2 [R]	66	68	124±16	123±15	74±1	79±1	25±1	27±1
Sub3 [R]	72	79	119±16	111±18	70±1	70±1	22±1	19±1
Sub4 [T]	68	76	119±16	115±16	71±1	75±1	12±1	22±0

Note: Seat angle is the planar angle at the midpoint between left and right greater trochanters between the crank axis and the handlebar; Hip angle is thigh segment relative to trunk segment; Anterior trunk lean is trunk segment relative to the vertical; Anterior pelvis tilt is relative to the vertical. [T]: triathlete; [R]: road cyclist.

The pattern of the hip joint angle during a pedal cycle was similar between seat conditions. The shift in the magnitude in some individuals appears to be adjusted by re-positioning the pelvis and/or trunk as mentioned above. Knee joint flexion angle was not affected by the seat position conditions in any of the study participants. Both the magnitude and the pattern of the knee angle were preserved with the seat position modification. This finding is consistent with Savelberg et al. (2003). Although no change was observed, this consistency is likely due to the adjustments occurring both proximal and distal to the knee. A few individuals exhibited altered ankle range of motion across the seat conditions. However, each study participant had a consistent pattern in the ankle angle data. There was a tendency of greater anterior tilt of the pedal while riding in the TRI seat position. DeGrood et al. (1994) observed a similar trend among elite cyclists where the pedal was tilted more posteriorly, especially during the downstroke when the seat tube was shallow (67°) compared to when it was steep (80°), although the resultant pedal force did not differ. They suggested that the pedal forces are affected if the intersegmental orientation among the leg, seat tube, crank and the pedal is modified since any changes in intersegmental orientation can affect the length-tension relationship of the working muscles. However, in the current study, as the intersegmental orientation changed (Figure 1) it is possible that pedal forces could be changed as a result of the different seat positions.

CONCLUSION: Changing the fore-aft seat position affected certain segmental and joint kinematics. The more extended position at the hip region was accomplished by vertically aligning the seat with the crank axis; however, the strategy used to obtain the extension appears to vary among the individuals. The changes in kinematics imply that seat position modification likely influences the working length of the leg muscles and the pedal forces. To obtain the comprehensive understanding of the effects of the seat position in cycling mechanics, other measurements, such as pedal forces, joint moments and electromyography should be considered in future investigations.

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ISBS 2010

Oral Session 07

Sprinting

FOOT PLANTING TECHNIQUES WHEN SPRINTING AT CURVES

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The paper presents the research on foot and shin positioning at the moment of contacting the ground when sprinting at a curve. The results attained allow for expanding the knowledge of foot performance when running. The study revealed that experienced sprinters, when sprinting at curves at maximum speed, have their right and left foot toes turned outwards, the right foot being turned more due to the need to resist centrifugal force.

KEYWORDS: sprint, sprinting at curves, foot.

INTRODUCTION: The technique of curve sprinting has repeatedly appeared in the focus of scientific research due to its exceptional importance for achieving outstanding performance (e.g. Greene, 1985; Jindrich, Besier & Lloyd, 2006). Yet, the peculiarities of foot plant and its further kinematics have not received the researchers' attention very often; in most cases, this technical element was studied from the lateral position (Umarov, Primakov & Tyupa, 1992; Umarov, 2000). Meanwhile, the characteristics of post-impact body motion is determined not only by the impulses they had before contact, but also by where the forces are applied. The human foot has a complex structure and possesses a number of physical characteristics not taken into consideration in most studies. A widespread assumption when analyzing curve sprinting is the uniaxiality of ankle-joint. This considerably simplifies the analysis of athlete's movements and the results achieved, as any change in foot plant is likely to cause considerable changes in the kinematics of foot during the support phase (and not only in sagittal plane), thus determining the preferred technique and sprinting efficiency on the whole.

It is impossible to evaluate the efficiency foot positioning in straight and curve sprinting while relying only on the results of the experiment in which the subject(s) would plant their feet on the race track in a various manner. In this case, not only would the mechanical characteristics of the athlete's locomotor system come into play, but also the neuropsychic characteristics. Foot plant is, by all means, a skill, and ruining the skill with the aim of forming a new one in the shortest possible time can certainly affect the efficiency of movement.

The aim of this research was to identify the peculiarities of experienced athletes' foot plant when running turns at maximum speed.

METHOD: We used 2-D video analysis of sprinters' foot and shin positions in frontal plane at the moment of their touching the track. The video was taken from the front by JVC GR-D379E camcorder with 50 Hz frame rate. The camcorder was positioned at a tangent to the curve at a predicted point of foot plant.

The following foot and shin characteristics at plant instant were determined: foot angle (α); sole angle (β); shin angle (γ) (Figure 1). Point 1 marks the middle of the horizontal line segment between the intersection of vertical tangent and the inside of the shin and the outermost point of the shin. Point 2 marks the middle of the horizontal line going from malleolus medialis into the outermost points of the shin. Point 3 is the distal point of the spike shoe toe. Point 4 and Point 5 mark the middles of holes for the rear spikes of the shoe. Segments 6-2, 2-7, and 4-8 are vertical; Segment 5-8 is a horizontal one.

The population of the study was comprised of six male sprinters (age: 19.3$\pm$1.51 years; height: 1.83$\pm$0.05 m; body mass: 76.2$\pm$5.2 kg; 100-m race performance: 11.1$\pm$0.3 s; 200-m race performance: 22.9$\pm$0.4 s).

Videotaping was done on the fourth lane, at the following points of the curve: 1) 30 meters from the beginning of the curve with starting line at the beginning of the curve ("entry point"); 2) 66 meters from the beginning of the curve with starting line marked 25 meters ahead of 200-m race starting line ("middle point"); 3) 96 meters from the beginning of the curve with starting line marked 55 meters from of 200-m race starting line ("exit point"); and also at the 30-th meter of a 40-meter straight path.

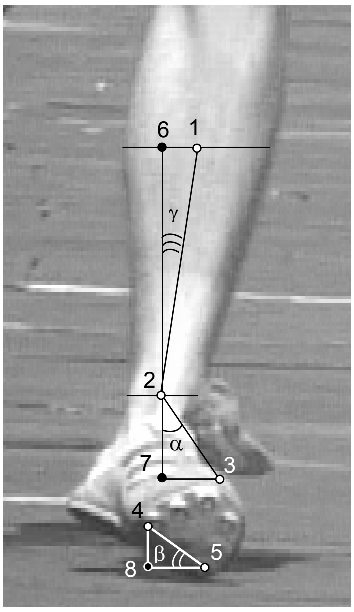

Figure 1. Calculation of foot angle (α), sole angle (β), and shin angle (γ)

Data collection lasted for two days. Three out of six subjects had to run 40-meter distance twice (for videotape of the right and left foot), following the procedure outlined above: at the entry and middle points on the first day, and at the exit point and on straight path on the second day. The other three subjects performed the order in reverse: on straight path and at the exit point on the first day, and at the middle and entry points on the second day). All began running at high start.
The statistical significance of sample data variation was estimated with single-factor analysis of variance (ANOVA).

RESULTS: The main results are presented in Table 1.

Table 1. Mean±S.D. angles of foot, shin, and sole at the plant moment when running various parts of the curve and during straight path running

Angle	Right	Left	p value
	Entry		
Foot	33.7 ± 2.8	20.1 ± 3.6	<0.001
Shin	3.3 ± 1.4	20.9 ± 2.8	<0.001
Sole	8.6 ± 5.4	28.0 ± 5.8	<0.001
	Middle		
Foot	29.3 ± 3.7	19.6 ± 5.0	<0.001
Shin	2.0 ± 1.4	22.6 ± 2.0	<0.001
Sole	17.3 ± 3.9	24.1 ± 5.6	<0.001
	Exit		
Foot	31.6 ± 4.2	17.5 ± 4.2	<0.001
Shin	1.4 ± 0.8	24.1 ± 2.0	<0.001
Sole	18.4 ± 2.8	30.4 ± 5.2	<0.001
	Straight path		
Foot	29.5 ± 4.2	28.8 ± 1.9	>0.05
Shin	8.7 ± 2.3	8.5 ± 1.9	>0.05
Sole	35.7 ±10.3	31.4 ± 7.1	>0.05

Angular values for right and left foot on all segments of the curve differed considerably, while remained similar for straight sprinting. It was found out that left (inside) foot in all three videotaping points was planted with the toe turned towards the inside of the lane, although the foot angle at the moment of plant when running on the curve was smaller than during straight sprinting ($p<0.001$). Note that beginning from sprinter's entering the curve, the foot angle gradually diminished (the significant difference between entry and exit point values was $p<0.001$).
The right (outside) foot was turned to the right. The angle of the right foot exceeded that of the left; in the middle of the curve, this value was practically the same as for straight sprinting ($p>0.05$). At the entry and exit points the right foot angles were slightly bigger than those observed during straight path sprinting ($p<0.01$).
Most of the subjects demonstrated that at the moment of touchdown, especially at the entry and middle points of the curve, their right (outside) shin was positioned almost vertically; at the same time, the left shin was visibly tilted with its proximal end aimed towards the center of the curve. When running on the straight, the angle of the tilt was the same for both right and left shins.
The table shows that the angle of the right foot sole is clearly smaller than that of the left one on every segment of the curve. It also proves to be clearly smaller on every segment of the curve compared to straight running. The right foot plant is the flattest when entering the curve (the difference between the sole angles at entry and exit points is statistically significant); yet, in the middle and at the curve exit point these angles are practically the same ($p>0.05$). The differences between the angles of the left sole when running on the curve and straight running reached statistically significant values once only, i.e. in the middle of the curve.

DISCUSSION: Biomechanical evaluation of curve sprinting determined clear differences in foot and shin positions at the moment of contacting the ground. When doing the curve, the left foot is planted with the toe pointed inside the curve at a sole angle slightly smaller than when running on the straight path; the toe of the right foot is more turned outwards, and the foot plant is flatter. It is speculated that such foot positioning allows for more efficient resistance to centrifugal force due to the work of shin muscles.
At the moment of foot plant, the angles of the left and right shins at any point of the curve are considerably different: the left shin's proximal end is much tilted inwards the curve; the right shin is only slightly tilted inwards, and in some cases retains its vertical position. Thus, it is wrong to think that during sprinting on the curve "the whole body" tilts towards its center (Umarov, Primakov & Tiupa, 1992; Umarov, 2000); at least, the tilt angles of different body parts might be very different. No doubt, these different positions of left and right shins result from the resistance to centrifugal force, though the mechanism of this counteraction is not that obvious compared to when the right (outside) foot is planted outwards.

CONCLUSION: The results of the study reveal that the need to counteract the centrifugal force results in sprinter's planting his/her right foot outwards more than the left one, and in a much flatter manner. This helps the athlete to keep balance more effectively.
The further developments in the study of curve sprinting techniques might require the following: 1) 3-D video analysis of movements produced by foot, shin and the whole locomotor system of sprinters when running both on the curve and on a straight path; 2) study of elite sprinters' foot and shin kinematics in support phase; 3) biomechanical modeling of curve sprinting technique employing various foot planting techniques.

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KINEMATIC ASPECTS OF BLOCK PHASE TECHNIQUE IN SPRINTING

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This study investigated kinematic aspects of block phase technique during the sprint start and their relationships with performance amongst a heterogeneous group of 16 sprinters. Lower limb kinematics in the 'set' position were not associated with block phase performance (average horizontal external power). During block exit a greater rear leg push, in particular from the hip, appeared important for performance. The front leg extended in a proximal-to-distal fashion, with more rapid hip extension again facilitating performance. Striving to achieve higher levels of block phase performance did not appear to negatively affect the first flight phase or the configuration of the sprinters at first touchdown. Sprinters should therefore be encouraged to maximise hip extensions in the blocks and use their rear leg drive to achieve a powerful block exit.

KEYWORDS: athletics, performance, sprint start, track and field.

INTRODUCTION: Since starting blocks were introduced to the sprint events in athletics in 1928-29, the block phase has been the subject of numerous descriptive and experimental biomechanical studies. A large volume of this research has focussed on 'set' position technique, reporting considerable inter-subject variation and weak relationships between 'set' position kinematics and performance (Mero, 1988). However, there exists limited research which has quantitatively determined how the lower limb joint angles change during the block phase once a sprinter reacts to the starter's gun and moves from the 'set' position, and it is not clear how these kinematics influence performance.

External kinetics during the block phase have been widely documented (Payne & Blader, 1971; Baumann, 1976; Mero, 1988; van Coppenolle et al., 1989; Lemaire & Robertson, 1990). Where force has been measured separately on each block face, the higher block exit velocities of better starters have been attributed to an increase in force generation with the rear leg (Payne & Blader, 1971; van Coppenolle et al., 1989; Lemaire & Robertson, 1990). However, due to the dearth of descriptive kinematic data from the block phase, the actual techniques which more successful starters use to achieve these higher levels of performance remain unknown.

A further issue that must be considered is that the block phase is not a 'stand-alone' part of a sprint, and that simply striving to maximise block phase performance could affect performance during subsequent phases of a sprint. Therefore it is also important to investigate whether achieving high levels of block phase performance could potentially inhibit technique and performance during the next stance phase. The aim of this study was to identify the lower limb angular kinematics associated with higher levels of block phase performance and to assess any relationships between block phase performance and kinematics at the first touchdown on the track.

METHODS: Sixteen male sprinters (mean ± *s*: age = 21 ± 5 years, height = 1.78 ± 0.05 m, mass = 74.4 ± 8.3 kg) ranging in ability from world-class (fastest 100 m PB = 9.98 s) to university-level (slowest 100 m PB = 11.6 s; Table 1) provided informed consent for high-speed video data to be collected from one of their training sessions. For 13 of the sprinters, data were collected indoors just prior to the competition phase of the indoor season. For the remaining three sprinters, data were collected outdoors during the competition phase of the outdoor season. Each sprinter completed three maximal effort sprints to 30 m, commencing from starting blocks which were adjusted to their preference. At all sessions, a high-speed digital video camera (Motion Pro®, HS-1, Redlake, USA; 200 Hz) was mounted on a tripod, and images were collected at a resolution of 1280 × 1024 pixels. Indoors, an area of 2.00 m

horizontally by 1.60 m vertically was calibrated at the centre of the running lane inside a 2.50 m wide field of view (restricted by only being able to position the camera 8.00 m away from the lane of interest). Outdoors, the camera was positioned 40.00 m from the lane centre, and an area of 3.50 m horizontally by 1.60 m vertically was calibrated inside a 4.00 m wide field of view. Due to limitations with the camera set-up, rear foot data for Sprinter A in the 'set' position were unavailable, and this sprinter was thus removed from the analysis when variables reliant upon rear foot data from the early block phase were required.
The video clips were digitised using a zoom factor of 2 (Peak Motus®, v.8.5, Vicon®, USA). Eighteen points (vertex, C7, shoulder, elbow, wrist, third metacarpal, hip, knee, ankle and second metatarsal-phalangeal joint centres) were manually digitised from one frame prior to movement onset until 10 frames after first stance touchdown. Following backward replication of the first frame 10 times to alleviate any potential for endpoint error, the data were filtered using a fourth-order Butterworth digital filter with cut-off frequencies determined individually for each displacement time-history via residual analysis (16 to 28 Hz). Joint angles at specific events (e.g. 'set' position, block exit) and peak joint angular velocities during specific phases (e.g. rear leg push, total push phase) were identified. Block exit velocities were calculated from the first derivative of a linear polynomial fitted through the raw CM displacement data (calculated using segmental inertia data from de Leva, 1996) from the flight phase immediately following block exit (Salo & Scarborough, 2006). The change in kinetic energy during the block phase was calculated from these velocity data, and was divided by the duration of the total push phase to determine average horizontal external block power (hereafter termed block performance) as a measure of performance (Bezodis et al., 2008). Block performance and all linear displacements were normalised to account for body size (Hof, 1996). Mean values for each sprinter were calculated and where appropriate, Pearson's correlations were run to determine the relationship between specific variables using these mean data from each of the 16 sprinters.

RESULTS AND DISCUSSION: Across all 16 sprinters, a strong negative relationship ($r = -0.71$, $p < 0.01$) existed between 100 m PB time and block performance, highlighting that better overall sprinters were also typically better starters. There were exceptions to this (most noticeably Sprinters D and G; Table 1), which indicated that block phase technique should be compared against block phase performance, not performance measures including subsequent phases of a sprint. Joint extension ranges of motion during block contact varied considerably between sprinters (Table 1). Whilst all sprinters extended their front hip joint over the greatest range of all the front leg joints, the largest rear leg extension typically occurred at the hip, but also at the knee for two sprinters. Rear hip range of motion during rear block contact was moderately correlated with block performance ($r = 0.44$, $p = 0.09$), although the rear hip angle at block exit was more strongly correlated ($r = 0.58$, $p < 0.05$) suggesting greater rear hip extension through the higher end of its range of motion may be important for performance. In addition to this apparent trend for rear hip extension to facilitate performance, a greater push duration with the rear leg (as a % of total push phase duration) was also associated with higher levels of block performance ($r = 0.49$, $p = 0.06$). These results reinforced previous suggestions (Payne & Blader, 1971; van Coppenolle et al., 1989; Lemaire & Robertson, 1990) regarding the importance of the rear leg push against the blocks, and it appeared that extension of the rear hip plays an important role in this.

Table 1. Ability level (100 m personal best (PB) in seconds), block phase performance and lower limb joint angle ranges of motion (°) during respective block contacts (mean values).

	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O	P
100 m PB*	9.98	10.22	10.35	10.51	10.53	10.70	10.90	11.10	11.19	11.2	11.2	11.3	11.3	11.55	11.6	11.6
Block power†	6.46	6.00	6.28	5.50	6.91	6.64	4.05	5.57	6.04	6.01	5.15	4.99	4.73	4.49	4.20	4.45
Rear hip	44	31	53	26	26	53	25	13	20	50	17	37	16	24	36	27
Rear knee	18	21	28	8	10	21	25	19	14	17	14	23	17	16	27	14
Rear ankle	n/a	19	29	14	19	15	24	35	22	9	14	27	17	15	28	0
Front hip	117	110	124	118	107	125	113	95	103	130	117	103	113	107	112	111
Front knee	76	79	77	66	66	75	74	73	74	69	87	66	74	57	85	74
Front ankle	32	35	49	22	47	30	44	45	25	34	34	37	50	22	41	21

*100 m PB times reported to the nearest 0.1 s are hand timed. †Normalised horizontal block power.

The relationships between peak joint extension angular velocities and block performance were generally weak, with only two correlation coefficients exceeding ± 0.40. The correlation between block phase performance and peak front hip angular velocity was r = 0.56 ($p < 0.05$), and at the rear hip was r = 0.43 ($p = 0.09$), suggesting an importance associated with the rate of extension of the front, and potentially the rear, hip joint. The temporal pattern of peak leg joint angular velocities (Figure 1) revealed that all 16 sprinters showed a rear leg sequencing of knee-hip-ankle. In contrast, all sprinters (except Sprinter C) exhibited a proximal-to-distal hip-knee-ankle extension pattern with the front leg. This proximal-to-distal pattern was unsurprising, since it is commonly associated with power demanding tasks due to the action of the biarticular muscles facilitating a transfer of power down the leg (Jacobs & van Ingen Schenau, 1992). However, such a strategy was not used when extending the rear leg, which could be due to the knee joint starting from a more extended angle in the 'set' position (group range = 95° to 122° compared to 78° to 95° in the front leg). The rear knee therefore could not extend for long, limiting its force producing capability. It appears that in the rear leg, hip joint extension is of major importance, due not only to the aforementioned relationships with performance, but also the increased time over which it is extending during rear block contact. The limited knee extension may be a strategy designed to reduce the rear block contact time after the initial extension of the hip, allowing the leg to swing through in preparation for the first ground contact on the track.

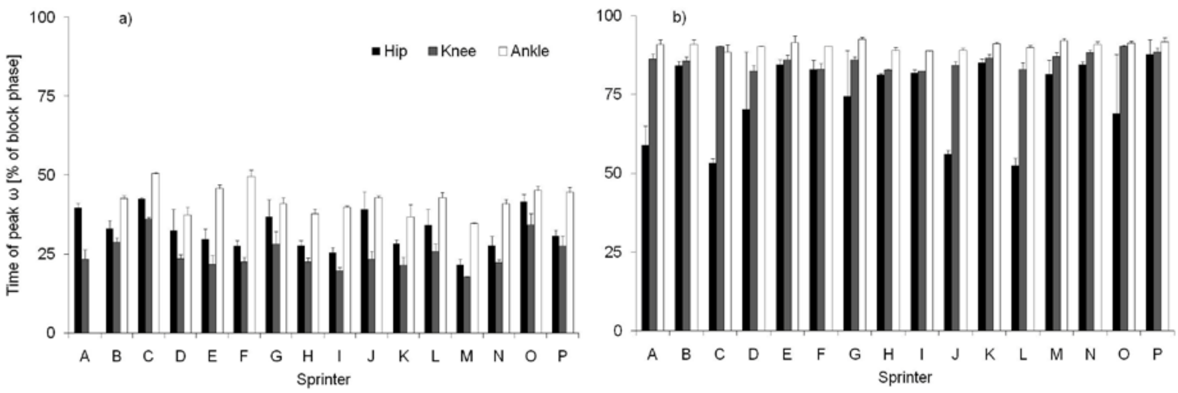

Figure 1. Timing of peak extension angular velocities at a) the rear leg and b) the front leg.

In the 'set' position, no lower limb or trunk angles were correlated with block performance above a strength of ± 0.24 ($p > 0.05$). This confirmed previous suggestions regarding the lack of an 'optimal' block positioning that is applicable for all sprinters. Smaller rear knee and hip angles in the 'set' position were correlated with increases in rear foot push duration (r = -0.59, $p < 0.05$ and r = -0.57, $p < 0.05$, respectively). These increased durations subsequently appeared to allow the rear hip to extend over a greater range during rear block contact and also reach greater extension angular velocities (respective correlations with rear foot push duration were r = 0.83, $p < 0.001$ and r = 0.80, $p < 0.001$). The use of personalised block settings therefore appears paramount, and specific adjustments could be made to address certain deficiencies in technique, such as reducing rear knee and hip angles if an increased push with the rear leg is required.

Beyond block exit, a large inter-subject range of stance leg joint angles existed at first touchdown (Table 2). These stance leg configurations at touchdown affected touchdown distance (the horizontal distance between the CM and the stance toe at touchdown, with negative values representative of the toe behind the CM; Table 2). Touchdown distance can have a considerable effect on a sprinter's ability to generate propulsive force during stance, since the CM must be rotated further in front of the stance foot prior to leg extension for this extension to propel the sprinter in a more favourable horizontal direction (Jacobs & van Ingen Schenau, 1992; Bezodis et al., 2008). However, whilst levels of block performance did not appear to affect the subsequent flight duration (r = 0.19, $p = 0.48$), potentially favourable trends existed between block performance and (normalised) step length (r = 0.41, $p = 0.12$)

and (normalised) touchdown distance (r = -0.42, p = 0.10). This suggests that striving to increase block performance does not appear to inhibit subsequent performance in a sprint, since sprinters tended to take longer steps and land in a better position at touchdown without major increases in flight duration, although the causality of this cannot be determined.

Table 2. First flight duration (ms), step length, touchdown distance and stance leg angles (°) at touchdown.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O	P
Flight time	85	102	90	63	82	68	62	50	78	80	57	70	27	70	122	67
Step length*	1.14	1.18	1.24	1.12	1.11	1.03	1.03	1.04	1.09	1.17	1.10	1.05	1.15	1.03	1.12	1.02
T/d distance*	-0.23	-0.28	-0.16	-0.21	-0.30	-0.29	-0.22	-0.18	-0.23	-0.20	-0.12	-0.18	-0.03	-0.23	-0.23	-0.17
Hip angle	96	95	98	99	103	99	111	73	91	103	91	96	79	97	98	86
Knee angle	102	112	93	100	99	102	94	93	100	99	101	106	89	115	110	103
Ankle angle	112	105	91	98	98	102	100	83	95	94	95	96	83	100	96	93

* Normalised step length and touchdown (T/d) distance using the convention of Hof (1996)

CONCLUSION: The results of this study highlighted the role of extension of the leg joints during the block phase. For all sprinters, all three joints in the front leg extended over at least 20° in a proximal-to-distal extension pattern. Whilst the rear leg joints extended over a smaller range, this rear leg extension, in particular at the hip joint, was associated with higher levels of block phase performance. Due to the differing strategy adopted with each leg in the blocks it is possible that different legs may be more suited to either the front or back block. Coaches should adjust block settings on an individual basis, and make specific changes if deficiencies in block phase technique are identified (e.g. if the rear leg push is short or weak, the blocks should be adjusted to facilitate slightly more flexed rear knee and hip angles in the 'set' position). As higher levels of block performance were not subsequently associated with any potential decrements in technique at the onset of the first stance phase, sprinters should be encouraged to maximise extension with both hips during the block phase in an attempt to achieve maximal horizontal external power production.

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KINEMATIC ANALYSIS OF HURDLE CLEARANCE OF 60-M HURDLES IN ELITE HURDLE SPRINTERS DURING WORLD INDOOR CHAMPIONSHIPS 2010

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KEYWORDS: Athletics, competition analysis, sprinting.

INTRODUCTION: Previous studies have examined the biomechanical variables of sprint hurdling of world-class athletes (Mero, & Luhtanen, 1986; McDonald, & Dapena, 1991). However, less is known about the factors that differentiate the performance among elite hurdle sprinters. Salo, Grimshaw, & Marar, (1997) compared international and national/county level female hurdlers and found that better hurdlers use greater take-off distance, which enables lower take-off angle and greater horizontal take-off velocity. Our previous analysis between international level hurdlers and decathletes revealed an opposite pattern of greater take-off distance for decathletes than for hurdlers (Kuitunen, Palazzi, Poon, & Peltola, 2007). The present study aims to examine the possible differences in hurdle clearance between different level of elite hurdle sprinters.

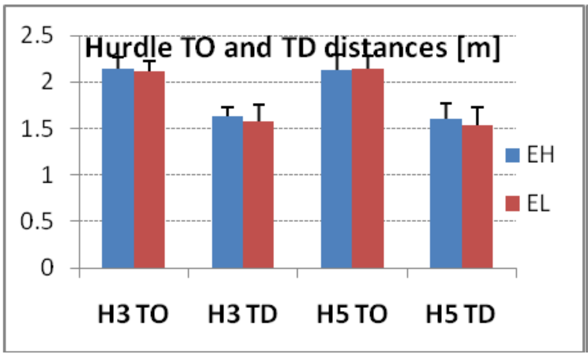
METHOD: Data was collected during the 13th World Indoor Championships in Athletics (March 12-14, 2010, Doha, Qatar). Semifinals and final of the men's 60-m hurdle races were recorded with four video cameras. Two panning high speed cameras (300 frames/sec) were placed above the spectator stands adjacent to 3rd (H3) and 5th (H5) hurdles for analyzing the take-off (TO, the instant of foot leaving the ground before the hurdle) and touchdown (TD, the instant of foot touching the ground after the hurdle) moments to each hurdle. Footages from the both cameras were synchronized by the starting gun light signal. In addition, two stationary video cameras (50 fps) were set perpendicular to the track at H3 and H5 for analyzing the TO and TD distances to the hurdle.
The hurdle sprinters (n=22) were divided into two groups according to their race performance: elite high group (EH; <7.7s) (n=10) and elite low group (EL; >7.7s) (n=12). The fastest race for each athlete was selected for analysis. Intermediate and interval times between the hurdles (between two consecutive TDs) and hurdle clearance times (from TO to TD) were calculated from the high speed video data and the TO and TD distances (distance from the tip of the shoe to the hurdle at the TO and TD, respectively) were determined from the footage of stationary cameras. Hurdle clearance velocity was determined by dividing the total hurdle clearance distance (TO distance + TD distance) by the hurdle clearance time. Known track marks were used as calibration for distance measures. Official results and reaction times were provided by Seiko (official timing for IAAF). T-test for independent samples was used to compare the differences between the groups and P<.05 was set at the level of significance. Data is presented as group means (± SD).

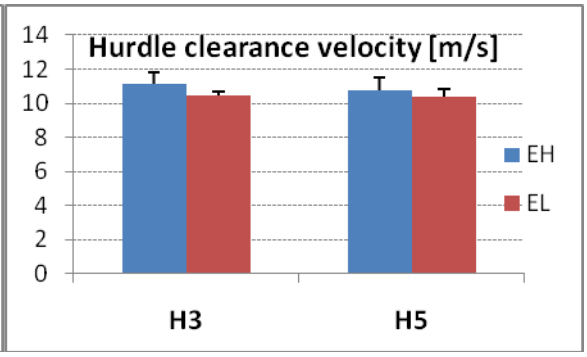
RESULTS: The mean race results for EG and IG were 7.55±0.13 and 7.82±0.06s (P<.001), respectively. The EG athletes demonstrated significantly faster reaction times (0.155±0.018 vs. 0.189±0.031s, P<.01) as well as intermediate and interval times as compared to the IG athletes (P<.05 - .001) (Table 1).
No differences were found in the hurdle clearance times between the groups (EG 0.35±0.03 vs. IG 0.36±0.02s). However, it showed a significant correlation to race result within the entire subject pool (r=.63, P<.01). The TO and TD distances were neither significantly different between the groups, although the EG showed a trend for slightly greater TD distance (1.63±0.10 vs. 1.58±0.17m for H3 and 1.61±0.16 vs. 1.53±0.20m for H5, respectively) (Fig. 1). On the other hand, EG demonstrated greater hurdle clearance velocity than IG for H3 (11.12±0.66 vs. 10.44±0.25 m•s^{-1}, P<.01) (Fig. 2). The hurdle clearance velocity was also found to correlate significantly with the race result (r=-.80, P<.001 for H3 and r=-.61, P<.01 for H5).

Table 1. Time analysis for 60m hurdle races for the two athlete groups.

Group	Time [s]	H1	H2	H3	H4	H5	Result
EH	Intermediate	2.58[2]	3.63[2]	4.64[3]	5.65[3]	6.67[3]	7.55[3]
	Hurdle	0.35	0.35	0.34	0.34	0.35	
	Interval		1.05[1]	1.02[3]	1.01[3]	1.02[1]	0.87[3]
EL	Intermediate	2.66	3.73	4.79	5.84	6.90	7.82
	Hurdle	0.37	0.36	0.35	0.35	0.35	
	Interval		1.07	1.06	1.05	1.05	0.92

[1,2,3] P<.05, P<.01, P<.001, significantly different between the groups





Figures 1 (left) and 2 (right). Take-off (TO) and touchdown (TD) distances to the hurdle (left) and hurdle clearance velocity (right) for the hurdles 3 (H3) and 5 (H5) for the elite high (EH) and elite low (EL) groups.

DISCUSSION: The present findings suggest that neither hurdle clearance time nor TO and TD distances to the hurdle differ between top level hurdle sprinters. This is different from the previous studies using hurdle sprinters of lower performance level (Kuitunen, Palazzi, Poon, & Peltola, 2007; Salo, Grimshaw, & Marar, 1997). Most likely the differences in hurdling performance among elite hurdle sprinters are fairly small as indicated by the present data. However, hurdle clearance velocity seems to play a role in differentiating the race performance among elite hurdle sprinters. This emphasizes the importance of achieving high horizontal velocity in between the hurdles and maintaining it during the hurdle clearance (Salo, Grimshaw, & Marar, 1997).

CONCLUSION: Differences in hurdle clearance are very small among world-class hurdle sprinters and the main difference is likely related to achieving and maintaining high horizontal velocity for the hurdle clearance.

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PERFORMANCE DETERMINING FACTORS IN ELITE SPRINTERS DURING SPRINT START AND TWO FOLLOWING SUCCESSIVE SUPPORTS

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KEYWORDS: sprint start, kinematics, kinetics

INTRODUCTION: Sprint start out of the blocks and successive acceleration are technically challenging as the athlete goes from a bended to a forward leaning position. Therefore, the body center of mass (COM) has to be accelerated forward and upwards. Optimal sprinting performance relies on attaining maximal forward acceleration. However, adequate vertical acceleration must be generated to reach sufficient height to prepare for the following step (Weyand, 2000). Horizontal acceleration is mainly determined by the horizontal ground reaction force that will affect sprint velocity and therefore final sprint performance (Mero, 1988). Kinematics and kinetics of the start action and maximal sprinting were intensively studied; however little is known on the transition from the set position to the running position during the first two strides. This study aims to identify the factors in the start action as well as in the first and second contact after block clearance that determine sprinting performance in terms of speed and acceleration.

METHODS: Sprint starts were analyzed in 19 elite athletes (8 males and 11 females) of the Royal Belgian Athletics Association. All participants gave their written informed consent to participate. After an individual warm-up and a static trial, each athlete completed three, ten meter sprints out of starting blocks on a tartan surface. Best out of three trials was included for further processing.
Kinematics were assessed by three-dimensional motion analysis (Vicon, Oxford Metrics, UK). Data were collected using 12 MX3 cameras (250Hz) which were positioned around the start action (starting block and the first two strides). A full body marker placement protocol consisting of 66 markers was used. An instrumented starting block with load cells mounted on the back of each starting block registered bilaterally the horizontal force-time characteristics during the entire starting action. Ground reaction forces were measured during the first two contacts using 2 force plates (Kistler, 1000Hz) embedded in the track. After initial processing in Nexus, further data analysis was performed in Opensim (Delp, 2007). By means of inverse kinematics, lower limb, trunk and upper limb kinematics were calculated based on a scaled musculoskeletal model comprising 29 DOF. Hip flexion/extension, hip ab/adduction, knee flexion/extension and ankle plantar and dorsiflexion were included in the current analysis. For the two contacts, only kinematics of the support limb were reported. Based on the force signal, impulses were calculated and normalized with respect to the athlete’s body weight. Correlation analyses were performed using Statistica 8.0. Significance level was set at 0.05.

RESULTS AND DISCUSSION: During set-position before gunshot, no significant correlations were observed between the kinematics and acceleration or speed at take off from the block.

During the remaining start action, positive correlation between knee extension of front leg and ankle plantar flexion of front and rear leg were found with block acceleration (Correlations respectively $r = 0.58$, $r = 0.53$ and $r = 0.50$). No correlation with hip extension was retained. Hip adduction of the front leg was negatively correlated with acceleration (r = -

0.50). Excessive adduction of front leg at end of block contact impairs the transfer of the body weight towards the rear leg therefore hindering the acceleration.

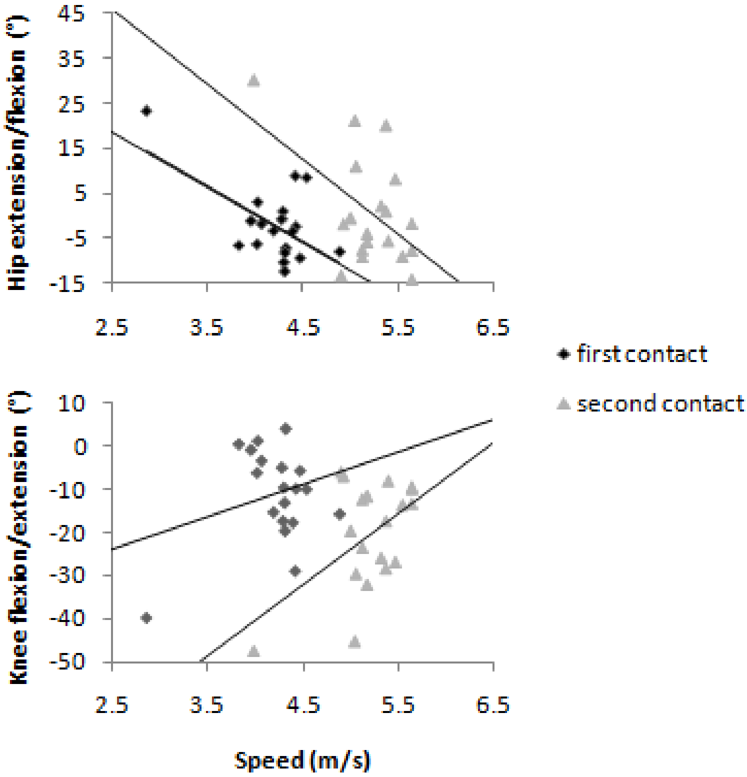

Figure 1. Scatterplot of hip flexion/extension and knee flexion/extension at toe off of first and second contact.

During first contact: initial hip flexion correlated positively with the breaking impulse (r = 0.48) and negatively with speed at the end of first contact (r = -0.46). During the remainder of support, increased hip flexion would generate a higher braking impulse (r = 0.48 for the whole group and r = 0.65 for women). At toe off, hip extension correlated positively with speed at takeoff. During second contact, breaking impulse correlated with knee extension (r = 0.82) at initial contact. In male runners, a strong correlation was found between hip extension and breaking impulse (r = 0.97), whereas in the full group, takeoff speed is positively influenced by both hip and knee extension. At toe off, hip and knee extension as well as plantar flexion in the ankle joint were positively correlated with takeoff speed (resp. r = 0.78; 0.78 and 0.75).

CONCLUSION: Different kinematic factors influence the braking impulse for first (hip flexion) and second (hip and knee extension) initial contact. No correlations were found with the propelling impulse for both contacts. Speed at toe off will be affected by similar kinematic parameters (extension of lower limb joints).

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CHANGES IN SPLIT VELOCITIES DURING SPRINT PERFORMANCE DEVELOPMENT

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Sprint times and split velocities are invaluable measures for coaches and athletes monitoring sprint training and performance development. This study analysed sprint times and 10 m split velocities as performance of three developing athletes developed over a five week training period. All significantly improved their 60 m sprint times over the training period ($p < 0.05$). Sprint performance developed individually with a tendency for maximal velocities to increase early in the training period and start and acceleration velocities later. All athletes' performances fluctuated between weeks, possibly due to a period of experimental learning in their process of skill development. This study will inform further analysis of the kinematic and kinetic parameters determining velocity, with the aim of identifying the key variables responsible for these changes.

KEYWORDS: Athletics, speed, split times

INTRODUCTION: The fundamental aim of sprint training is to decrease sprint time over a set distance, commonly, 60, 100 or 200 m. For an athlete to improve they must either increase sprint velocity or be able to sustain maximum velocity for a longer distance. Consequently, in addition to sprint times, it is valuable for coaches and athletes to have feedback of split times and average split velocities throughout the sprint. Such information will enable identification of weaknesses in an athlete's performance and aid the monitoring and evaluation of training progress and interventions.

Previous studies have concentrated on cross-sectional analyses of velocities (e.g. Mann & Herman, 1985; Ae et al., 1994); few have used a longitudinal approach to monitor sprint performance development. As part ongoing research into sprinting biomechanics, this longitudinal study focused on the early stages of training and in particular on how an athlete's velocity profile changes as technique develops and performance improves. Such insights will support sprint training, particularly in developing athletes and will potentially enhance training efficiency (Young et al., 2001). Additionally, velocity analysis over an entire 60 m run, and encompassing each sprint phase, will provide insights into where and how velocities vary during individual and series sprints. Such information has not been readily available due to the lack of availability of the necessary measurement technology and the phase-specific nature of previous sprint studies. Further understanding of how the velocity profile changes as sprint performance develops will also inform subsequent analysis of kinematic and kinetic parameters determining velocity, with the aim of identifying the key variables responsible for changes. The aim of this study was to examine how split velocities varied as performances developed for a group of non-sprint specific athletes over a five week training period. Non-sprint specific athletes were used to ensure performance developments were made over this time.

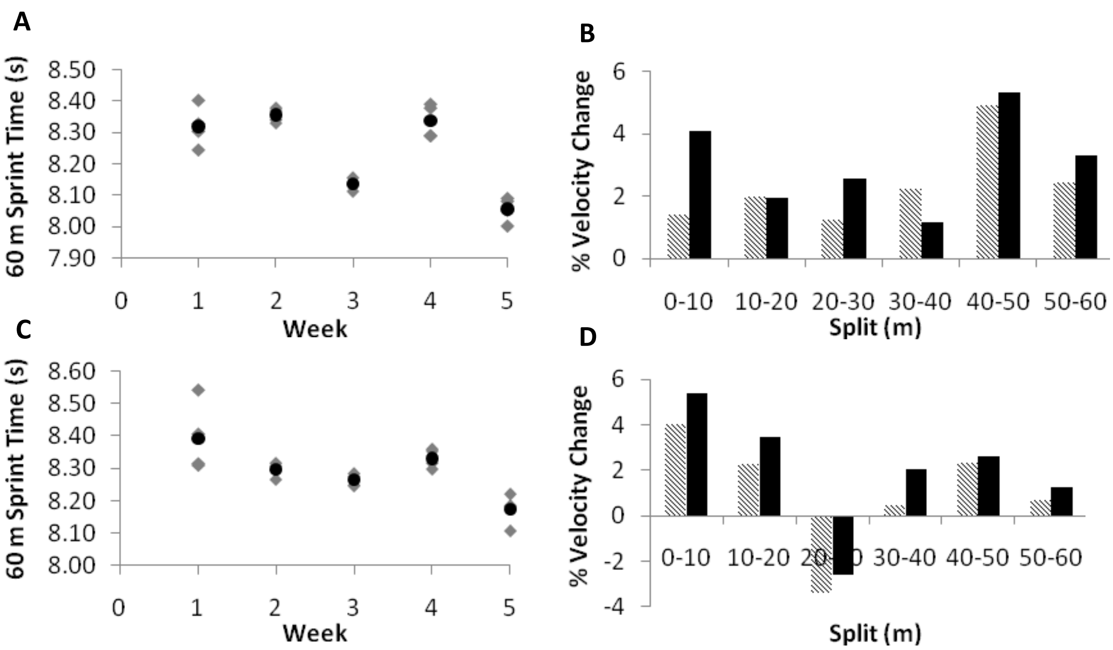
METHOD: **Collection:** Three male recreational athletes (Athlete 1: height 1.81 m, mass 87.7 kg, age 19 yrs, 60 m personal best (PB) 8.09 s; Athlete 2: 1.75 m, 75.6 kg, 20 yrs, 8.10 s; Athlete 3: 1.70 m, 71.1 kg, 19 yrs, 8.65 s) gave written informed consent to participate in the study. The athletes had no history of serious or recent injuries and were fit for the duration of data collection. They trained at an indoor athletics centre twice a week for five weeks, completing five 60 m sprints per session against fellow developing athletes. During each session, separated by at least three days, the athletes performed a similar warm up and wore the same running shoes. A ceiling mounted light gate timing system (PLG, Cheng et al., 2010) was used to record six '10 m splits' for each trial and the athletes were provided with their sprint times following every trial. The 10 m splits corresponded approximately with different phases of sprint performance: Start (0-10 m), acceleration (10-20 & 20-30 m), high velocity (30-40 m) and maximum velocity (40-50 m & 50-60 m). A block start was initiated by

an audible signal (hooter). Rest between each trial was never less than 5 minutes, minimising the effects of fatigue.
Sprint times for 60 m and 10 m split times were output by the PLG system and average velocities for each 10 m were calculated from the split times. The athletes' four fastest sprints per week were selected for all analyses. Mean ± SD sprint times and split velocities were calculated for each of the five weeks. Significant differences in sprint times and split velocities were identified using Repeated Measures ANOVA with Bonferroni correction for repeated measures. The level of significance (*p*) was set *a priori* to 0.05.

RESULTS: Over the five week training period the athletes PB 60 m times improved significantly (Table 1). The average 60 m time decreased significantly ($p < 0.05$) over the training period (Table 1 & Figure 1). Differences in velocity were evident and percentage change between Week 1 and Weeks 3 and 5 are displayed (Figure 1). Table 2 presents the statistically significant mean differences in split velocities over the five weeks.

Table 1. 60 m PB and mean (±SD) of the four fastest sprints for each week and the mean difference between weeks 1-5 (* $p < 0.05$)

Athlete		Week 1	2	3	4	5	Mean Difference
1	Best 60 m time (s)	8.24	8.33	8.11	8.29	8.00	
	Mean 60 m time (s)	8.32	8.36	8.14	8.34	8.06	0.26*
	SD	0.06	0.02	0.02	0.05	0.04	
2	Best 60 m time (s)	8.31	8.27	8.25	8.30	8.11	
	Mean 60 m time (s)	8.39	8.30	8.26	8.33	8.17	0.22*
	SD	0.11	0.02	0.02	0.03	0.05	
3	Best 60 m time (s)	8.95	8.75	8.67	8.66	8.66	
	Mean 60 m time (s)	8.99	8.80	8.74	8.78	8.69	0.30*
	SD	0.04	0.06	0.08	0.08	0.02	



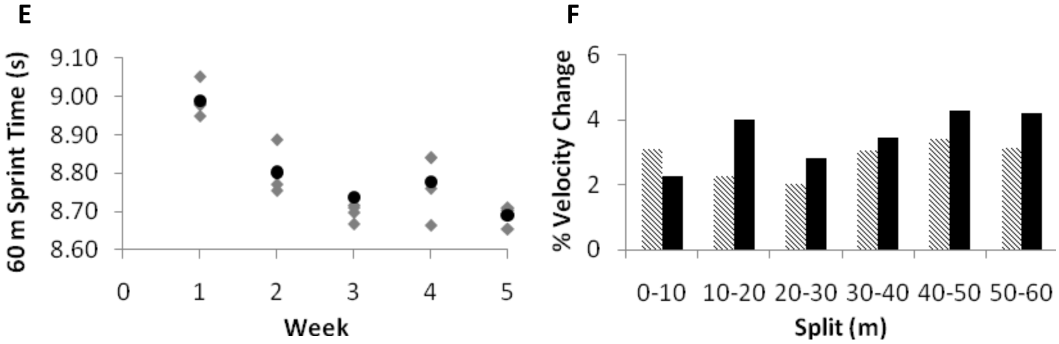


Figure 1. A, C, E: 60 m sprint times (◆) and mean per week (●). B, D, F: Percentage velocity change between split velocities in the first, mid and last weeks. The shaded bars indicate change weeks 1 - 3 and black weeks 1 - 5. Results are for the best four sprints from each week for athletes 1 (A & B), 2 (C & D) and 3 (E & F).

Table 2. Significant (*p* < 0.05) mean differences (a-b) in split velocities between weeks.

	Athlete 1			Athlete 2			Athlete 3		
Phase	Week	Week	(a-b)	Week	Week	(a-b)	Week	Week	(a-b)
0-10	5	1	0.18	5	1	0.24	3	1	0.13
		2	0.17						
		4	0.20						
10-20	4	3	-0.31	5	1	0.27	5	1	0.30
		5	-0.31		4	0.18	5	3	0.13
20-30	5	1	0.22						
		2	0.41						
		4	0.21						
30-40	4	3	-0.28	5	2	0.21	5	1	0.28
		5	-0.19		4	0.26		2	0.26
40-50	3	1	0.41	3	4	0.18	4	2	0.13
		2	0.33	5	1	0.22	5	1	0.35
		4	0.33		4	0.21		2	0.21
	5	1	0.47						
		2	0.39						
		4	0.38						
50-60	5	1	0.29				3	1	0.24
		2	0.36				4	1	0.25
							5	1	0.35

DISCUSSION: This study used case studies of three developing athletes to monitor sprint times and split velocities over a five week training period. The aim was to identify specific patterns in improvement of sprint time and velocity, informing the coach and further research. The changes in sprint times (Table 1) indicate the athletes improved their sprinting abilities over the ten session training period. Figure 1 illustrates how 60 m sprint times decreased due to increases in velocity across weeks 1 – 5. Figure 1 also reveals that performance improvement did not occur between every week and this is particularly evident where Week 4 times decrease from the times of Week 3 in all three cases. However, despite these apparently undulating sprint times, performance between the beginning, middle and end of the training period improved. The rate of this improvement is particularly consistent in athletes 1 and 2. Further analysis is required to understand why sprint performance varied to the extent it did here but this may be explained by the athletes' stage of learning. Collectively, these analyses indicate recreational athletes' sprint performance can be improved over a five week period but variations should be expected within this time (Newell et al., 2001). It is of interest to both coaches and researchers to understand how an athlete develops sprinting ability and identification of split velocity changes (Figure 1 & Table 2) over the training period will add insight to this.

Although faster average velocities were found for most splits as each athlete's performance improved, only the significantly different ($p < 0.05$) velocities are discussed here.

The development in Athlete 1's sprint performance by Week 3 can be attributed to a significantly higher maximum velocity (40-50 m). The further improvement by Week 5 is due to a faster start (0-10 m) and continued development in the maximum velocity achieved (40-

50 m) and maintained (50-60 m). The slow performances of Weeks 2 and 4 are due to slower acceleration (10-20 m) and high velocity (30-40 m) phases.
Athlete 2 achieved no significant improvements in average split velocities until the final week of training. Improvements in the start (0-10 m), acceleration (10-20 m) and maximum velocity (40-50 m) phases were then evident. Surprisingly, as Athlete 2 developed his sprinting ability, average velocity over the 20-30 m split decreased throughout the training period. Further analysis is required here but this may be indicative of changes in technique in acceleration affecting the transition into maximum velocity.
Athlete 3 showed an initial improvement in his start (0-10 m) which was followed by decrements between Weeks 3 and 5. This may indicate that Athlete 3 required a longer period of training to refine this particular skill. Conversely, he may have altered his starting technique to improve his acceleration phase, which did increase in velocity from Weeks 3 - 5 (0.13 s). Athlete 3 also increased average velocity in the high and maximum velocity phases. He improved his ability to maintain maximum velocity earlier in the training period (Week 3) than he was able to significantly increase the maximum (40-50 m, Week 4) and high (30-40 m, Week 5) velocities.
In summary, all three athletes improved their sprinting ability at different rates and through development at different phases of the sprint. Initial improvements in performances were largely due to improved split velocities at high (30-40 m) and maximal (40-50 m) velocity whereas decreased sprint times later in the period were due to slight increases in maximal velocity but also large improvements in the start (0-10 m) and/or early acceleration (10-20 m) phases. Athletes 1 and 2 showed no significant improvement in their start phase until the last week of training, perhaps suggesting the start is a difficult phase to improve and a greater level of technical skill may require more practice than the later more familiar phases. All three athletes demonstrated a trend towards increased maximum velocity and maintenance of velocity as performance improved.
Variations in sprint performance, particularly evident in Week 4 for all subjects, could be an essential part of improving sprint technique. Such patterns may be due to the athlete entering an experimental time while learning and developing their technique (Wilson et al., 2008). The changes reported in this study were possible without any technical feedback available to the athlete. Development of a quantitative method of technical feedback of variables determined as key to improving sprint performance may enable athletes to develop their technique beyond the rate and level that their own proprioceptive ability can achieve. This may be particularly true for novice and developing athletes, or those without regular access to a high level of coaching expertise.

CONCLUSION: Collectively, these analyses indicate a recreational athlete's sprint performance can be improved over a five week period, but variations should be expected and may be necessary within this time. Further research of this type is required to enhance understanding of sprint performance development, extending knowledge for coaches aiming to enhance training specificity and efficiency.

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ISBS 2010

Oral Session 08

Jumping

New Investigator Award

FORCE, VELOCITY, AND POWER ADAPTATIONS IN RESPONSE TO A PERIODIZED PLYOMETRIC TRAINING PROGRAM

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This study assessed kinetic, kinematic and temporal adaptations to the countermovement jump in response to a 6 week periodized plyometric training program. Twenty recreationally active women were randomly assigned to a plyometric training or control group. Testing consisted of 3 maximal countermovement jumps on a force platform prior to and after the six weeks of training. A two-way repeated measures ANOVA was used to assess differences between pre- and post-testing sessions and between groups. Post-test eccentric and concentric velocity, power and jump height were significantly greater ($p < 0.05$) in the training group. Periodized plyometric training is effective for increasing jump height, and augmentations are likely due to enhanced eccentric velocity.

KEY WORDS: ground reaction forces, stretch-shortening cycle, program design

INTRODUCTION: Jumping ability is important for many athletic events. Plyometric training subjects have consistently demonstrated improvements in jumping performance (Markovic, 2007; De Villarreal et al., 2009). However, the specific kinetic, kinematic and temporal characteristics associated with improved jump height remain equivocal.
Jump height has been shown to be correlated with power and velocity (Dowling and Vamos, 1993) and is increased by countermovement jump training (Cormie et al., 2009). Specifically, power and velocity curve analyses have revealed higher values throughout the concentric and eccentric phases, as a result of countermovement jump training (Cormie et al., 2009). To better understand the jump height augmentation of the countermovement jump, the eccentric phase must be investigated.
Studies have shown the influential role of acutely reducing the duration of the amortization phase and increasing the eccentric speed, to increase the subsequent performance (Ham et al., 2007; Moran and Wallace, 2007; Wilson and Flanagan, 2008). More specifically, increased countermovement depth and eccentric velocity have been shown to be a positive adaptation of plyometric training (Cormie et al., 2009). This previous study, however, failed to include women and to implement Matveyev's model of periodization into the training program (Matveyev, 1966).
Therefore the purpose of this study was to explore the kinetic, kinematic, and temporal adaptations in response to a 6 week periodized plyometric training program for women. By quantifying kinetic, kinematic and temporal factors resulting from plyometric training, practitioners will gain insight into how increases in performance are manifested. This information may lead to the manipulation of training methods to enhance jump performance.

METHODS: Twenty subjects were randomly assigned to either a non-training control or a plyometric training group. Ten women served as controls (mean ± SD; age = 19.50 ± 1.18 years; height = 1.63 ± 0.065 m; body mass = 61.70 ± 9.90 kg). Ten women served as training subjects (mean ± SD; age = 19.00 ± 0.82 years; height = 1.68 ± 0.067 m; body mass = 62.72 ± 9.22 kg). All subjects provided informed written consent and the study was approved by the institution review board.
Warm-up prior to the testing and training sessions consisted of three minutes of low intensity cycling on an ergometer followed by dynamic stretching exercises as well as 5

countermovement jumps of increasing intensity. Five minutes of rest were provided prior to beginning the testing and training.

The plyometric training group trained twice per week with 48- to 96- hours recovery between training sessions for 6 weeks. The program was periodized by decreasing volume and increasing intensity based on previous recommendations (Potach and Chu, 2008). Volume was reduced from 100 foot contacts early in the program to 60 foot contacts upon cessation of training. Intensity was based on previous research examining kinetic variables of various plyometric exercises (Jensen and Ebben, 2007). Subjects rested 15 seconds between single jumps and 30 seconds between sets which was based on the previous recommendations of work to rest ratios of at least 1:5(Read and Cisar, 2001; Potach and Chu, 2008).

All training and control group subjects refrained from physical activity during the 6 week training period, as determined by subject activity logs. Subjects participated in one pre- and one post-training testing session consisting of three maximum countermovement jumps with arm swing performed on a force platform (BP6001200, AMTI, Watertown, MA, USA). Vertical ground reaction force (GRF) was sampled at 1000Hz, real time displayed, and saved with the use of computer software (BioAnalysis 3.1; AMTI, Inc.) for later analysis. Velocity was determined by subtracting body weight from the force-time curve, dividing by body mass, and integrating with respect to time using the trapezoidal rule (Dowling and Vamos, 1993). Power was determined by multiplying the GRF without body weight by velocity. The beginning and end of the eccentric phase was identified with methods previously used (Hori et al., 2009). The time in air (TIA) method was used for calculating jump height derived from the force platform (Aragón–Vargas, 2000). Concentric peak force, velocity and peak power were determined as the maximum values obtained during the concentric phase of the countermovement jump. Eccentric peak force was determined as the maximum force during the eccentric phase. Peak eccentric velocity was determined as lowest velocity value during the eccentric phase. Temporal variables such as eccentric and concentric duration were also assessed. These variables have been chosen due to their previously assessed importance to jump performance (Dowling and Vamos, 1993; Cormie et al., 2009).

All data were evaluated with SPSS 17.0 (IBM Corp., Chicago, IL, USA) using a two-way repeated measures ANOVA to assess the interaction and main effects for jump height, eccentric and concentric duration, time to peak force and take off, peak eccentric and concentric force and velocity, and peak power between pre- and post-testing sessions as well as between training and control groups. Significant interactions and main effects were further analyzed using paired-samples *t*-tests to identify specific differences in the outcome and performance variables pre- and post-testing. Assumptions for linearity of statistics were tested and met. Statistical power (*d)* and effect size (η_p^2) are reported. The a priori alpha level was set at $p \leq 0.05$.

RESULTS: Analysis of variance revealed significant interactions for jump height ($p \leq 0.001$, $d = 0.992$, $\eta_p^2 = 0.55$). There were no other significant interactions for the variables assessed ($p > 0.05$). However, results revealed significant main effects for pre- and post- testing: jump height ($p \leq 0.001$, $d = 1.00$, $\eta_p^2 = 0.74$), peak eccentric velocity ($p = 0.006$, $d = 0.85$, $\eta_p^2 = 0.36$), peak concentric velocity ($p = 0.022$, $d = 0.66$, $\eta_p^2 = 0.26$), peak power ($p = 0.005$, $d = 0.85$, $\eta_p^2 = 0.36$) and body mass ($p = 0.023$, $d = 0.65$, $\eta_p^2 = 0.25$). Paired-samples *t*-tests for the training group revealed that all of the aforementioned variables were different from pre-testing values with the exception of body mass ($p \leq 0.05$). Paired-samples *t*-tests for the control group revealed that all post-testing values were not different from pre-testing values ($p > 0.05$). The mean, standard deviation, and percent increase for the outcome and performance variables of the training group are shown in Table 1. Because follow up analyses of the control group variables revealed no significant differences, pre- and post-testing, only training group data are reported.

Table 1. Training group pre- and post-training outcome and performance variables (n=10).

Outcome and performance variables	Pre-training		Post-training		Percent Increase (%)
	Mean (SD)	CV(%)	Mean (SD)	CV(%)	
Jump Height(m)	0.24 (0.04)	16.67	0.29 (0.03)	10.34	20.83*
Ecc Duration(s)	0.62 (0.13)	20.97	0.64 (0.10)	15.63	3.23
Con Duration(s)	0.33 (0.06)	18.18	0.34 (0.06)	17.65	3.03
Time to Pk Force(s)	0.81 (0.18)	22.22	0.83 (0.13)	15.66	2.47
Time to Take Off(s)	0.95 (0.18)	18.95	0.98 (0.13)	13.27	3.16
Pk Ecc Force(N)	502.07 (144.54)	28.79	505.24 (202.36)	40.05	0.63
Pk Con Force(N)	702.04 (147.54)	20.96	731.54 (107.84)	14.74	4.2
Pk Ecc Velocity(m/s)	−0.78 (0.14)	−17.95	−0.99 (0.12)	−12.12	−26.92*
PkConVelocity (m/s)	2.36 (0.19)	8.05	2.50 (0.19)	7.6	5.93*
Pk Power (W)	134.37 (308.70)	22.98	1500.83 (167.37)	11.15	11.72*

Ecc = Eccentric; Con = Concentric; Pk = Peak; *Significant difference ($p \leq 0.05$) between pre-test and post-test

DISCUSSION: Six weeks of periodized plyometric training enhanced peak eccentric and concentric velocity, peak power and jump height by 26.92%, 5.93%, 11.72%, and 20.83%, respectively. Investigation of the eccentric phase adaptations provide insight into the underlying mechanisms responsible for the increased jump height.

Peak eccentric velocity was the only eccentric variable affected by the periodized plyometric training program. Research suggests that acutely decreasing ground contact or eccentric time will lead to greater return of elastic energy and thus enhanced performance (Wilson and Flanagan, 2008). The present study revealed that while eccentric velocity improved, eccentric duration was unaffected by training, but performance was enhanced. Other temporal variables including concentric duration, time to peak force, and time to take off, were also unaffected by plyometric training.

The increase in eccentric velocity augmented the peak concentric velocity and power and consequently, jump height, but not force. Previous research revealed similar increases in eccentric and concentric velocity, power and no change in peak force in response to countermovement jump training (Cormie et al., 2009). Despite similar changes to countermovement jump kinetics and kinematics, jump height was increased by 20.83% in the present study compared to 13.5% in the study using countermovement jump training (Cormie et al., 2009).The present greater increase in peak eccentric velocity and ultimately jump height could have been due to progressing the intensity of the exercises (depth jumps) that result in higher negative velocities compared to the countermovement jump. Thus, other factors such as training status, plyometric exercise type, intensity, and volume likely contributed to the present improvement in jump height.

The 5.0cm increase in jump height for women is higher than the average 2.82cm increase found by meta-analyses (De Villarreal et al., 2009). Training status has been suggested to be a factor in plyometric training adaptations (Markovic, 2007) however, De Villarreal et al., suggested that plyometric training gains seem to be independent of fitness level (De Villarreal et al., 2009).

The type of plyometric exercises incorporated into a program ultimately dictates the exercise intensity (Jensen and Ebben, 2007; Potach and Chu, 2008). Thus, plyometric exercise type can be manipulated to follow the periodization model. In fact, the periodized program used in the present study progressed from low intensity exercises such as line and cone hops to depth jumps and bounding movements, which have been shown to be of high intensity (Jensen and Ebben, 2007). Furthermore, the magnitudes of the changes in jump height from the periodized program are unique considering the short duration of the training program.

The present study demonstrated that a 6 week program periodized from 100 to 60 foot contacts per session was sufficient for jump height adaptations, compared to previous

recommendations for programs to be longer than 10 weeks consisting of 50 foot contacts per session (De Villarreal et al., 2009). Tapering or reducing the volume towards the end of the training program likely decreased fatigue, which influenced the significant improvement in jump height over few weeks (Matveyev, 1966).
It is noted that performing follow up analyses without a significant interaction may increase Type I error, however, previous research suggests that jumping performance is influenced by many kinetic, kinematic, and temporal variables, thus finding significant differences in individual predictor variables might be problematic (Dowling and Vamos, 1993).
Analysis of the eccentric phase and velocity of the countermovement jump provided insight into the manifestations of jump height increases in response to training. Additionally, the manipulation of program design variables and implementation of Matvyev's model of periodization influenced the eccentric and concentric phases and jump height during the countermovement jump.

CONCLUSION: Eccentric velocity was enhanced by 26.92% through periodized plyometric training which influenced the increases in concentric phase velocity, power and ultimately jump height by 5.93%, 11.72%, and 20.83%, respectively. This study highlights the importance of implementing periodization in plyometric program design by increasing intensity based on previous recommendations and decreasing volume by reducing the foot contacts over time.

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EFFECTS OF AGE, GENDER AND ACTIVITY LEVEL ON COUNTER-MOVEMENT JUMP PERFORMANCE AND VARIABILITY IN CHILDREN AND ADOLESCENTS

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The aim of this study was to investigate counter-movement jump performance and variability in a large population of children and adolescents with respect to age, gender, and activity level. 1835 subjects performed three counter-movement jumps with arms akimbo on a force platform. The subjects were divided into 6 age groups and three activity level groups. Jump height and maximum rate of force development were calculated for all jumps. The best trial out of three was considered for further calculations. Variability of both parameters was indicated by the coefficient of variation over three jumps. Both parameters increased with increasing age while their variability decreased. Boys jumped higher than girls. Regarding maximum rate of force development female subjects showed higher values. The active subjects jumped higher and with less variability than the sedentary group. Jump height and maximum rate of force development are good parameters to describe the development of jumping performance regarding age, gender and activity aspects. Due to the high variability of maximum force rate development, however, this parameter has to be interpreted with caution in subject-specific assessments.

KEYWORDS: jump height, maximum rate of force development, variability

INTRODUCTION: The counter-movement jump (CMJ) is a commonly used method in performance diagnostics to measure leg power and explosiveness. Most of the previously reported studies involved adult male subjects (Harmann et al., 1990, Marcovic et al., 2004). Only few studies focused on age and gender effects in jumping performance (Temfemo et al., 2008). The studies investigating gender differences in jumping performance are based on a small group of individuals and are limited to specific age groups such as prepubertal children (Temfemo et al., 2008), adolescents (Temfemo et al., 2008) or adults (Artega et al., 2000). No study was found dealing with gender-related differences in jumping performance in a large population during childhood and adolescence. Besides age and gender, the activity level might also be an important factor causing differences in jumping performance. Most previously published studies included only subjects highly experienced in jumping (Marcovic, et al., 2004, Vanezis & Lees, 2005). Data of non-active subjects are rare, although a comparison of different activity groups could provide important information about the application of vertical jumping in performance diagnostics for non-active subjects. To analyse age, gender or activity level related differences it could be helpful to consider parameters describing the variability of jumping performance. While commonly used parameters like maximum jump height or peak force only give information about mean or maximum performance, parameters describing the variability could be used to assess the stability of testing procedures and to explain differences between groups of subjects (Artega et al., 2000, Marcovic et al., 2004, Harrison & Gaffney, 2001). Harrison & Gaffney (2001) analysed coefficients of variation for different gender and age groups, but they only compared children with adults and the sample size was restricted to 42 subjects. Artega et al. (2000) and Marcovic et al. (2004) did not consider age or gender effects on variability and none of the previously reported studies compared variability of jumping performance in subjects with different activity levels. Thus, the aim of this study was to investigate the effect of age, gender and activity level on jump height and maximum rate of force development and the variability of these parameters in counter-movement jumps in a large population from childhood to adulthood.

METHODS: In the context of a comprehensive motor ability and motor skill survey, 1835 children and adolescents at the age between 4 and 17 years [10.6±3.9 yrs, 39.8±8.2 kg, 143.6±22.0 cm] were selected for this study. Regarding age, subjects were separated into six groups (4-5 yrs, 6-7 yrs, 8-9 yrs, 10-11 yrs, 12-14 yrs and 15-17 yrs). The activity level was determined using a detailed questionnaire. School sport, informal sport and organised sport activities of middle and high intensity were subsumed. An activity-index was established indicating the overall sport activities of each subject in hours per week. Based on these data the subjects were separated into three groups of activity level: 'active' (>8 hours sport activities per week), 'moderate' (3–8 hours sport activities per week) and 'sedentary' (<3 hours sport activities per week). All subjects performed three counter-movement jumps with arms akimbo on a force plate. The instruction was to jump as high as possible. Vertical ground reaction forces were measured with a sampling rate of 1000 Hz. Maximum jump height (h) was calculated by integrating the force-time curve and equalizing the kinetic and potential energy. The best out of three trials was taken for further calculations. Maximum rate of force development (RFD) was calculated for the best trial with respect to jump height. In addition, the variability of these parameters [cv-h, cv-RFD] was quantified using the coefficient of variation over three jumps calculated as:
cv (%) = [SD(trial 1-3)/mean(trial 1-3)] * 100.
A multivariate ANOVA with the factors age, gender and activity level was used to analyse differences between subgroups for all parameters. The level of significance for all tests was set a priori to 0.05.

RESULTS: Significant age effects were found for jump height and jump height variability (h: $p<0.001$, $\eta^2=0.468$, cv-h: $p<0.001$,$\eta^2=0.074$). Post hoc tests showed a significant increase in jump height with increasing age over all six age groups and a significant decrease in jump height variability until the age of 9 years. Significant age effects were also revealed for the maximum rate of force development (RFD: $p<0.001$, $\eta^2=0.238$), indicating a significant increase from 10 years of age onwards. No significant age effects were found for the coefficient of variation of the maximum rate of force development.
Regarding jump height and jump height variability, significant gender effects were found (h: $p<0.001$, $\eta^2=0.079$, cv-h: $p<0.001$, $\eta^2=0.016$), indicating greater jump height, but also higher variability for boys than for girls. The maximum rate of force development was significantly higher in the female subjects (RFD: $p<0.001$, $\eta^2=0.026$), however, no significant gender differences were found for the variability of maximum rate of force development.
A significant interaction between age and gender was observed regarding jump height, jump height variability and maximum rate of force development (h: $p<0.001$, $\eta^2=0.080$, cv-h: $p<0.001$, $\eta^2=0.014$, RFD: $p=0.003$, $\eta^2=0.010$). From the age of 12 years onwards boys increased their jump height more than girls, so that the difference of jump height between boys and girls changed from about 1.3 cm for the 4-11 years group to 9.4 cm for the group of 15-17 years old subjects. Jump height variability differed significantly between boys and girls in the younger age groups (4-5, 6-7 and 8-9 years), showing higher values for the boys. Regarding the maximum rate of force development, boys and girls at first developed similarly with a higher increase from 10 years of age onwards, while girls showed higher values than boys. This changed between the 12-14 and 15-17 years old subjects. From that age onwards the boys increased the maximum rate of force development much higher than the girls, so that significant differences no longer were found between boys and girls at the age of 15-17 years. No significant interaction between age and gender was observed for the variability of the maximum rate of force development.
Regarding the activity level of the subjects, significant differences between the three groups were revealed for the maximum jump height and jump height variability (h: $p=0.018$, $\eta^2=0.005$, cv-h: $p=0.020$, $\eta^2=0.005$). Post hoc tests showed significant differences only between the active and the sedentary subjects. While jump height increased with increasing activity level (sedentary: 20.0 cm, active: 21.2 cm), the jump height variability decreased with increasing activity level (sedentary: 9.1%, active: 6.8%). For the maximum rate of force

development and its variability no significant differences between the three activity groups were observed.

Table 1. Jump height, jump height variability, maximum rate of force development and variability of maximum rate of force development separated according to gender and age [mean (SD)]

		Age (years)					
	Sex	4-5 (n=292)	6-7 (n=277)	8-9 (n=274)	10-11 (n=258)	12-14 (n=423)	15-17 (n=311)
h (cm)	Boys (n=792)	13.5 (5.6)	16.8 (5.4)	19.2 (5.2)	21.7 (5.1)	25.1 (5.9)	31.8 (6.1)
	Girls (n=1043)	12.0 (4.1)	16.0 (4.4)	17.6 (3.2)	20.4 (4.5)	21.9 (4.7)	22.4 (4.8)
cv-h (%)	Boys (n=792)	16.4 (15.4)	11.7 (11.5)	11.6 (17.9)	6.9 (7.8)	6.3 (8.0)	5.9 (5.8)
	Girls (n=1043)	11.2 (10.3)	8.2 (8.1)	6.6 (7.2)	5.3 (4.7)	6.6 (8.5)	6.1 (7.6)
RFD (kN/s)	Boys (n=792)	3.5 (2.0)	3.6 (18)	4.3 (3.1)	5.3 (2.7)	6.6 (3.6)	8.9 (5.4)
	Girls (n=1043)	4.0 (2.3)	4.6 (25)	5.4 (3.4)	6.9 (3.4)	9.1 (4.7)	9.6 (4.9)
cv-RFD (%)	Boys (n=792)	23.1 (16.0)	25.8 (20.3)	22.9 (16.1)	22.5 (14.1)	20.0 (13.9)	20.5 (15.1)
	Girls (n=1043)	22.1 (13.5)	21.8 (15.0)	20.4 (13.6)	20.5 (11.8)	20.3 (12.7)	20.0 (11.9)

h: maximum jump height; cv-h: jump height variability; RFD: maximum rate of force development; cv-RFD: maximum rate of force development variability.

DISCUSSION: Nearly all investigated parameters in counter-movement jumps were affected by age. Jump height and the maximum rate of force development are good parameters to describe the development of explosive leg extension power during maturation. As expected, the highest jump height variability was observed in the 4-5 years old subjects and decreased with increasing age. The measured jump height variability of subjects from ten years of age onwards were in line with results of previously reported studies, which found jump height variability between 2.8% and 6.3% (Artega et al., 2000; Marcovic et al., 2004). For younger subjects aged between 4 and 9 years the results of the present study showed higher values of jump height variability between 16.4% and 11.6% for the boys and 11.2% and 6.6% for the girls. The variability of maximum rate of force development was much higher with 22.6% for the 4-5 years old subjects and barely decreased to 20.3% for the 15-17 years old subjects. Stokes (1985) reported magnitudes of variability inherent in biological systems of 10% to 15%, which emphasised the very high values for the variability of maximum rate of force development in all age groups. This leads to the conclusion that the maximum rate of force development is a very unstable parameter and may only be used for assessing group analyses. For individual assessments this parameter cannot be determined reliably enough in all investigated age groups.

The higher jump performance in boys is in line with previously reported studies and can be explained by different gender-related physical conditions (Temfemo et al., 2008; Harrison & Gaffney, 2001). Regarding maximum rate of force development boys developed lower values than girls. No other studies were found indicating these parameters between boys and girls during growth. In spite of the lower values in maximum rate of force development, boys jumped significantly higher than girls. The explosiveness in the eccentric phase obviously is not sufficient to guarantee high jumping performance. Boys are able to produce leg extensor muscle forces over a longer period of time leading to enhanced jumping heights. These abilities are better pronounced in boys than in girls.

Jump height was only slightly higher for boys until the age of 11 years. From the age of 12 years onwards the jump height difference between boys and girls increased continuously. This is in line with previously reported studies which showed increasing differences between boys and girls from the age of 11 or 12 years onwards (Temfemo et al., 2008). These gender-related developments can be explained by different pubertal changes in boys, which lead to an increase in leg lengths, leg muscle volumes, muscle forces and higher percentages of fast twitch muscle fibres (Temfemo et al., 2008). Regarding jump height variability, boys showed significantly higher values up to the age of 9 years. Along with the pubertal change the gender differences disappear. Regarding the rate of force development, significant differences between boys and girls were found indicating higher values for the girls. From the age of 12 years onwards boys assimilated to girls and no significant gender differences for the maximum rate of force development were found from this age onwards. No significant interaction was found for the other parameters. At the current stage of data

analysis this gender effect cannot be explained and has to be clarified by further detailed investigations.
The activity level only affected jump height and jump height variability. Active subjects jumped higher than the sedentary group and showed lower variability during consecutive performance. These results suggest that physical activity and sport experience enhances jumping performance. Although the differences are significant, they are small and might be interpreted as irrelevant. Sport specific and age-related analyses have to be performed to provide stronger interpretations of these findings. The higher variability for the sedentary group showed that subjects with no or little sport experience were not able to reach his or her maximum performance repeatedly in consecutive jumping. Therefore, several trials should be performed in jumping performance diagnostics to provide a greater chance of measuring a subject's actual best performance, specifically in inexperienced children.

CONCLUSION: The study shows a substantial enhancement of jumping performance during childhood and adolescence, both in males and females. This indicates a high sensitivity for developing jumping related abilities during this specific time-span. Specific training is supposed to be very successful in reaching high jumping performance. This is important as jumping performance is one of the basic abilities in many sports. The different gender-related developments indicate specifically high adaptations to jump power in boys starting with puberty. The variability of jumping performance is rather high in both sexes up to the age of 10 years. From that age on jumping performance diagnostics is able to produce valid and reliable data. Prior to that age, related data have to be interpreted with caution. Despite the relatively low variability in jumping performance it is necessary to provide at least three trials during diagnostic procedures. Otherwise it is not guaranteed to get sufficiently reliable data in jumping diagnostic testing. The data clearly show substantial enhancement of the maximum rate of force development in both sexes during maturation, indicating the enhancement of leg extension power. So this parameter can be used for studying group effects. Due to the large variability, however, individual data analyses have to be interpreted cautiously. Furthermore, the results support the necessity to use force plates in jumping performance diagnostics as the maximum rate of force development cannot be determined with any other methodology. For the interpretation of jumping performance data in children and adolescence both, the sport activity level and the area of sport activity have to be considered.

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EFFECTS OF EIGHT WEEKS PILATES TRAINING ON JUMP PERFORMANCE AND LIMITS OF STABILITY IN ELEMENTARY DANCERS

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KEYWORDS: postural stability, exercise training, core strengthening.

INTRODUCTION: Dance is not only a performing art, but also a highly rigorous athletic sport that is one of the most physically and mentally demanding athletic sports in the world (Shah, 2008). Dancers are a unique group of athletes in that they execute physically challenging movements while making them look beautiful and artistic. This performance ability requires a high level of fine motor control, strength, flexibility, and core stability. Pilates is a kind of conditioning which was used to exercise the body flexibility, muscles strengthen, and body alignment. It is suggested that the Pilates is good for joint flexiblity and core strengthing, however, limited study was conducted to evaluate its benefits on jump and postural stability. Therefore, the purpose of this study was to evaluate the effects of eight weeks Pilates exercise training on jump performance and limits of stability (LOS) in elementary dancers.

METHOD: Twenty-six elementary dancers equally and randomly assigned to experimental group (EG, age: 10.9 yrs, height: 147.5 cm, weight: 37.9 kg, dance experience: 3.5 yrs) or control group (CG, age: 11.2 yrs, height: 146.1 cm, weight: 36.1 kg, dance experience: 4.2 ± 0.8 yrs). All subjects received the same dance lessons as routine elementary curriculum but the experimental group underwent an extra Pilates mat exercises for 40 minutes, three times a week, for 8 weeks. A instructor who had 2 years of experience in Pilates mat exercises initiated the exercises.

Jump performance was evaluated by bilateral countermovement jump (CMJ) and squat jump (SJ) for maximal height on AMTI force plate with Noraxon TeleMyo 2400T G2 at sampling rate of 1500 Hz. Hands remained on the hips for the entire movement to eliminate any influence of arm swing. Jump technique was demonstrated by one of the investigators, followed by two sub-maximal attempts by the participant. Three jumps for maximal height, separated by 3 minutes rest, and the jump with the greatest height was subsequently used for data analysis.

Limits of stability (LOS) was evaluated by the Biodex Balance System. Subjects were tested bilaterally at two levels of difficulty: 2 and 8. To control for the learning effect and fatigue, the order of the tests was randomly assigned. The subject was instructed to start moving the cursor which accurately move the display toward the flashing target at eight different directions. The LOS score was calculated for each direction according to the percentage between the straight line distance to target and the number of samples. Therefore, more direct the path to the target and back to center, the higher score will be achieved.

All statistical procedures were performed by using SPSS Version for Windows 12 (Chicago, IL, USA). A mixed design, one-way ANCOVA was used to evaluate the difference between groups after training for each parameter. The statistic significance was set at $p < 0.05$.

RESULTS: The average CMJ jump heights in the EG and CG were 23±5 cm and 20±4 cm in pre-training, and changed to 23±4 cm and 19±4 cm after the eight-week period ($F = 10.66$, $p < 0.05$). The average SJ jump heights in the EG and CG were 20±5 cm and 18±3 cm in pre-training, and changed to 19±4 cm and 17±4 cm after the eight-week period ($F = 0.42$, $p > 0.05$). The overall LOS scores for the EG group at levels 2 and 8 pre- and post-training changed from 22.6±8.3% to 31.3±8.9% and from 44.6±5.2% to 56.1±4.7%, respectively. The overall LOS score for the CG at levels 2 and 8 in pre- and post-training changed from 26.3±10.5% to 27.3±11.2% and 43.3±4.3% to 56.3±5.2%, respectively. The results of the ANOVA for the overall and left direction LOS scores at level 2 (Figure 1) indicated a significant interaction between the trained/untrained groups × pre/post repeated measures with post-training scores higher than pre-training scores in the EG (F=11.06 & 4.5, $p < 0.05$).

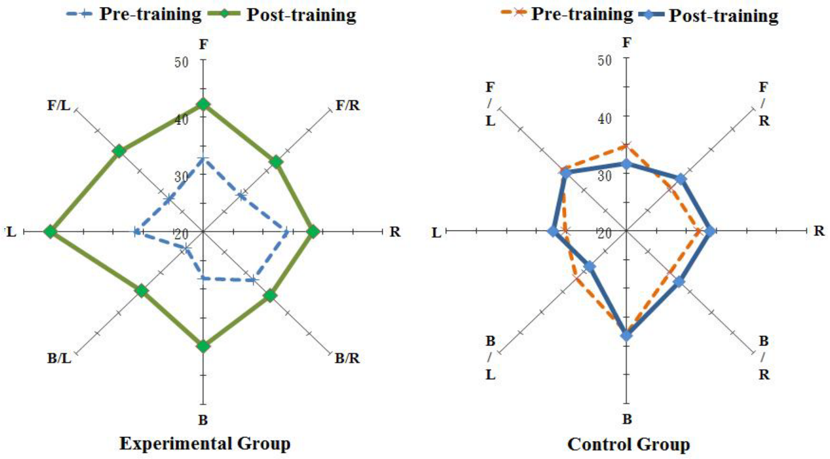

Figure 1. LOS scores in each direction between groups at level 2 before and after training.

DISCUSSION: This study demonstrated that young dancers who participate in 8 week Pilates training can improved the CMJ jump performance and total LOS performance. The CMJ is defined as a fast and powerful movement using an active eccentric contraction induces a powerful concentric contraction (stretch-shortening cycle, SSC). So it's a better indicator of neuromuscular coordination and control, were the SJ is a better indicator of explosive power. Previous study examining national-level gymnast using Pilates twice a week showed significant increases in jump height 16.2% (Hutchinson et al., 1998) which was in accordance with the finding of this study.

In addition. One recent study have indicated that Pilates training can enhances the control of trunk movement, and improves the jump neuromuscular coordination of movements, thus enhancing the overall jump performance (Lugo-Larcheveque et al., 2006). The improved LOS performance which demonstrated in this study and enhanced body movement which demonstrated in Lugo-Larcheveque et al., (2006) might suggest that Pilates training can strengthen the core muscles then improve the stability of the upper body and the lower extremity limb coordination.

CONCLUSION: This study demonstrated that eight weeks Pilates exercise can improve CMJ and total LOS performance in elementary dancers. Therefore, Pilates exercise is benefitial and should be implemented into elementary dance curriculum.

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ANALYSIS OF STABLE FLIGHT IN SKI JUMPING BASED ON PARAMETERS MEASURED WITH A WEARABLE SYSTEM

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Biomechanics analysis of the ski jump is highly required. Some parameters and their interrelations have been reported in previous research studies limited to few athletes. The generalization of these parameters to athletes of various levels and under training conditions should be assessed, since they have the potential to be used for daily evaluation. This study proposed a new 3D approach based on inertial sensors to evaluate relevant kinematic and aerodynamic parameters of stable flight phase. The proposed wearable system can easily be used for daily training. Aerodynamic forces and body segments 3D angles were extracted during the stable flight phase of 86 jumps. Then, their correlations with respect to distance as well as their interrelations were analyzed. Their combination expressed 55% of the total distance variance.

KEYWORDS: Ski jump, inertial sensors, daily training, angles and aerodynamics, stable flight

INTRODUCTION: In ski jumping, the performance, which is scored by the jump length and style points, is influenced by the athlete movement during all the phases of the jump (i.e., in-run, take-off, early-flight, stable flight and landing). Several studies were conducted to identify measurable parameters that could be related to the performance (Schwameder, 2008). For example, Arndt et al. (1995), Müller et al. (1996) and Schmölzer and Müller (2005) proposed kinematic (e.g., joint angles and V-opening angles) and aerodynamic (e.g., drag and lift forces) parameters during stable flight. Three-dimensional (3D) optical-based capture systems were used to measure these parameters. However, these systems have a small capture volume, need a constraintful calibration and require complex data post-processing. Therefore, these measureable parameters were mainly studied in the frame of specific research applications with few athletes. It is thus necessary to determine if these observations are generalizable to wider population under daily training conditions.
Recently, body worn inertial sensors were proposed to study ski jumping. These small and lightweight devices are not constrained to a limited capture volume and do not require complex post-processing. They are therefore very promising instrumentation for routine in-field measurement of movement in order to provide real-time feedback to the athletes and coaches. However, currently in ski jumping this technology has only been used to extract drag and lift forces (Oghi et al., 2008) and to determine timing events (Aminian et al., 2009).
The first objective of this study was to propose a new easy-to-use method based on body worn inertial sensors to measure the movement during the entire ski jump. This method automatically calculates the timing events, the orientation (3D) of main body segments during the complete jump and the drag and lift forces during stable flight. The second objective was to analyze the correlations of previously suggested parameters with the jump length, as well as their interrelations during the stable flight phase.

METHOD: The measurement system was composed of seven wireless inertial-based modules (Physilog®, BioAGM, CH). Five modules were attached to the athlete by an underwear suit. These modules were located on the sacrum, laterally at the middle length of the thighs and at the middle length of the tibia shaft. The two other modules were fixed on the skis behind the back fixation. Each module, weighting less than 100g, was composed of a 3D gyroscope (±1200 °/s), a 3D accelerometer (±10g) and an embedded datalogger recording the signals at 500 Hz.

32 male athletes (19±4 years old, 173± 8cm, 58±8 kg) from Swiss ski jumping team, including juniors, Nordic combined world-class and world-class athletes, were enrolled in this study. They were asked to perform up to three jumps with the proposed wearable system in K105 hill jump in Einsiedeln (Switzerland) during summer season. Before each jump a calibration procedure was realized to align the inertial modules with the body segments as proposed in Favre et al. (2009). Standard video camera, synchronized with the wearable system, was recorded at 25Hz. Starting platform height, wind conditions, distance and professional evaluation of style were collected for each jump. A score based on distance and starting platform height was calculated for each jump.
To measure body segments orientations and aerodynamic forces, the following steps were performed. First, some temporal features (i.e., take-off, beginning of stable flight, end of stable flight and landing impact) were detected by identifying specific features in the inertial signals (Aminian et al., 2008). Then, based on the acceleration signals measured during in-run phase, an initial orientation was defined for each body segment. An algorithm fusing the angular velocities, accelerations and aerodynamic constraints was then used to calculate the segments orientations until landing (Bortz, 1971). Finally, aerodynamic forces were evaluated during stable flight phase applying the first Newton law to the sacrum acceleration. Using the sacrum orientation, these forces were projected in the jumping hill reception frame X'Y'Z' (see Fig.1), where X' and Z' corresponded respectively to drag and lift forces.
To date, several parameters of the stable flight were analyzed (Arndt et al., 1995; Müller et al., 1996; Schmölzer and Müller, 2005). However for shake of consistency, only a subset was considered in this study. Moreover, some literature parameters were adapted because the wearable system did not measure the trajectory. These parameters are described below and in Fig. 1a and Fig. 1b. 1) Median of ski angle α with respect to horizontal in sagittal plane. This angle was used instead of the angle of attack. 2) Median of angle β between ski and shank in sagittal plane. 3) Median of angle γ between thigh and sacrum in sagittal plane. 4) Medians of V-angles V_f and V_t between skis in frontal and transverse planes. This decomposition allowed a 3D description of skis angles. 5) Medians and differences between 5th and 95th percentiles of sacrum drag and lift forces F_d and F_l. For the lower limbs, the values of the left and right sides were averaged:

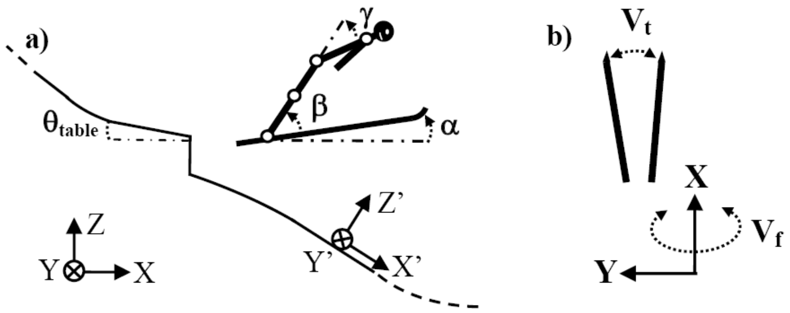

Figure 1. a) In sagittal plane, ski-horizontal angle (α), ski-shank angle (β), thigh-sacrum angle (γ) and representation of XYZ and X'Y'Z' frames corresponding respectively to jumping hill frame and jumping hill reception frame. b) V-angle between skis expressed in transverse plane (V_t) and in frontal plane (V_f).

Finally, the correlations between the extracted parameters and the distance score, as well as the correlations between the extracted parameters were evaluated based on Pearson linear method. P-values were also calculated to estimate the level of significance.

RESULTS: Body segments orientations and aerodynamic forces were reconstructed for a total of 86 jumps. Typical angles, as well as drag and lift forces curves are shown in Fig. 2. Correlations relative to distance score and between parameters are presented in Table 1. In addition, multiple regression analysis showed that the combination of all extracted parameters explained 55% (r=0.74) of the variance of distance score. For comparison, the angular parameters (α,β,γ,V_f and V_t) and forces parameters (F_d and F_l) explained respectively 42% (r=0.65) and 49% (r=0.70) of the variance of distance score.

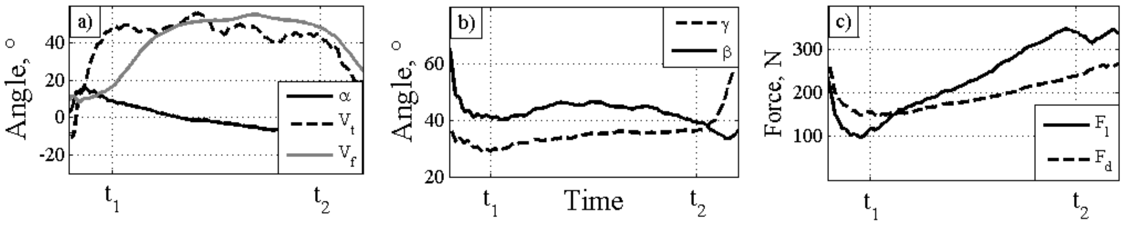

Figure 2. Kimatic and aerodynamic curves for a typical jump of a world-class athlete. t_1 and t_2 correpond to the beginning and to the end of the stable flight phase.

Table 1. Correlation coefficient r between each parameter and distance score

N=86		α	β	γ	V_f	V_t	F_d		F_l	
		MED	MED	MED	MED	MED	MED	PCTILE	MED	PCTILE
Distance score		**−0.45*** **	−0.24*	*NS*	**0.53*** **	0.26*	**0.31** **	**0.31** **	**0.62*** **	**0.57*** **
α	MED		**0.41*** **	*NS*	**-0.43*** **	*NS*	*NS*	*NS*	**-0.44*** **	**-0.60*** **
β	MED			*NS*	**-0.31** **	**-0.51*** **	*NS*	*NS*	*NS*	*NS*
γ	MED				*NS*	*NS*	*NS*	*NS*	*NS*	-0.25*
V_f	MED					**0.32** **	0.25*	*NS*	**0.62*** **	**0.41*** **
V_t	MED						**0.47*** **	**0.28** **	*NS*	NS
F_d	MED							*NS*	**0.46*** **	NS
	PCTILE								*NS*	**0.39*** **
F_l	MED									**0.54*** **

* $p < 0.05$; ** $p < 0.01$; *** $p < 0.001$; *NS* $p>0.05$

DISCUSSION: The inertial-based method presented in this study allowed measuring body segments orientations during the entire jump sequence and aerodynamics during stable flight. It also allowed an automatic extraction of angular and force parameters. Finally, this method did not alter the movement and could be used in daily training conditions. Generally the angles and forces curves (Fig. 2) obtained through this inertial-based system were in accordance to those in literature (Schmölzer and Müller, 2005). But we found slightly larger ski-shank angle curve β and V-opening angle curve V_t.

Regarding the aerodynamics, our results showed a good correlation between the distance score and the lift force F_l. Indeed, the correlations were positive for both the median value (r=0.62) and the increase (r=0.57) of F_l during stable flight. This could mirror an augmentation of velocity during flight and a proper aerodynamic posture. It was observed that the drag force F_d increased with lift force F_l. We hypothesized that this was due to a velocity increase. Schmölzer and Müller (2005) reported that the drag force augmentation has a minor negative effect on distance. In our study, both F_l and F_d were positively correlated with the distance. On the other hand, Müller et al. (1996) suggested to maximize the lift to drag force ratio. This suggestion was not supported in our study, where the lift to drag force ratio reported only a correlation of 0.36 with the distance. But, these statements should be considered with caution, since F_d and F_l were expressed in X'Y'Z' frame.

Regarding the body segments and skis kinematics, two parameters were highlighted with respect to distance: the ski angle α with repsect to horizontzal (r=-0.45) and the V-opening angle in the frontal plane V_f (r=0.53). When estimating the correlation by combining all angular parameters with respect to distance score, a total correlation of 0.65 was obtained. This indicates that the angle parameters were correlated to each other as illustrated in table 1. The ski-shank angle β was weakly correlated with the performance (r=-0.24), which differed a little from literature (Schmölzer and Müller, 2005; Schwameder, 2008). Except for a low correlation with lift force F_l, the sacrum-thigh angle γ did not show any significant correlation. Although the γ angle varied between athletes, it was not related to the performance. As mentioned by Müller et al. (1996) and Schwameder (2008), the V-technique allows reaching low ski-horizontal angle α (-0.43) and ski-shank angle β (r=-0.51). Furthermore, a high V_f, low α and low β suggested a more efficient flight posture, as indicated by their correlation with the lift force. Schwameder (2008) mentioned that the angle formed between skis was related to jump length. However, in our study, only V_f (influencing the distance between skis) showed an acceptable correlation with distance score. This was also reported by Arndt et al. (1995), since he found that the distance between skis was more relevant than the angle between them.

It is interesting to note that the parameters that athletes actually control during stable flight (α,β,γ,V_f and V_t) reported a good total correlation (r=0.65) with the distance score. The aerodynamic parameters (F_d and F_l) showed a similar correlation with distance (r=0.7). However, when considering all parameters the correlation only slightly increased (r=0.74). This suggests that force and angular parameters are correlated, as confirmed in Table 1. Indeed, drag and lift forces would be a part of the result of an aerodynamic posture.

CONCLUSION: In general, the correlations obtained on a large cohort during this study agreed with literature (Arndt et al., 1995; Schmölzer and Müller, 2005, Schwameder, 2008), where video cameras were used. Especially, the relevance of ski-horizontal angle α, V-opening angle V_f in frontal plane and lift force F_l was highlighted in regard to distance. But, ski-shank angle β and V-opening angle V_t were found weakly related to performance. It was the first time that skis kinematics during flight was described in 3D by considering α,V_t and V_f. Compared to previous studies, we had large differences of jumpers level. It is worth mentioning that by considering these nine parameters (all measured during the stable flight), 55% (r^2) of the whole variance of the distance score could be expressed. The analysis of the correlations between the proposed parameters improved the understanding of their interrelations. The inclusion of additional parameters (e.g., wind, in-run speed) would certainly improve the analysis. Finally, previous studies have shown that take-off and early-flight phases were also highly related to the performance (Arndt et al., 1995, Schwameder, 2008). Indeed they have a direct influence on the dynamics and kinematics reached in stable flight phase. The method proposed in this study could easily be extended to analyze the take-off and early-flight phases as well. Therefore, we can conclude that this inertial-based approach has a high potential for measuring and guiding ski jumpers during daily training.

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SHORT-TERM PLYOMETRIC TRAINING IMPROVES ALTERED NEUROMOTOR CONTROL DURING RUNNING AFTER CYCLING IN TRIATHLETES

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KEYWORDS: plyometrics, triathlon, neuromotor control

INTRODUCTION: Cycling has a direct negative effect on some highly-trained triathletes' ability to execute optimal neuromotor strategies specific to running (Chapman et al., 2008). The presence of altered neuromotor control when running off-the-bike has been associated with exercise-related leg pain (Chapman et al., 2010). Accordingly, identification of training interventions that could minimise this interference may aid in prevention of injury and augmentation of performance during running following cycling. Plyometric training is a specific form of strength training that has been reported to improve running economy by enhancing neuromuscular function (Paavolainen et al., 1999). The primary aim of this study was to examine the effect of plyometric training on triathletes neuromotor control and running economy in those in which neuromotor control is aberrant during running after cycling.

METHOD: A randomised control trial of an 8 week plyometric intervention program was conducted. Running economy and neuromotor control of fifteen well-trained triathletes (height, 175 ± 7 cm; weight, 65 ± 9 kg; VO_2max, 62 ± 6 ml.min^{-1}.kg^{-1}) were determined by measuring submaximal VO_2 and lower limb electromyography (EMG) for 4 min at 12 km.hr^{-1} during a control run (no prior cycling) and a run after 45 min of cycling (transition run). EMG data was collected from the tibialis anterior, gastrocnemius lateralis, rectus femoris and biceps femoris muscles of the right leg. Triathletes who exhibited altered neuromotor control in any muscle during running after cycling at baseline were included for further participation in the study. The criteria for altered neuromotor control was that the mean difference in EMG amplitude between control-and-transition runs exceeded 10% and that the 95% confidence intervals for EMG waveforms were not overlapping for ≥ 10% of the stride.

Subjects were randomly assigned to either a control or plyometric intervention program for a period of 8 weeks. Both groups continued their regular endurance training. Triathletes in the plyometric group also participated in 3 x 30 min plyometric training sessions per week. (see Table 1 for example of exercises and increasing difficulty of the program over the 8 weeks).

Table 1. Plyometric program (Weeks 1 & 8 given as an example of progression of program).

Week	1		6-8		
Session	1	2	1	2	3
Squat	2 x 6		5 x 8		
Countermovement jumps	1 x 6		3 x 6		
Knee lifts (technical)	1 x 20		3 x 20		
Ankle jumps	1 x 10		3 x 10		
Hamstring Curls	1 x 10		3 x 10		
Alternate leg bounds		1 x 10		6 x 10	4 x 10
Skip for height		1 x 30m		4 x 30m	5 x 20m
Single-leg ankle jumps		1 x 20m		4 x 20m	
Continuous hurdle jumps					5 x 5
Scissor jumps for height					5 x 8

Primary outcome measures of neuromotor control were (i) the EMG waveform; (ii) mean EMG amplitude; (iii) coefficient of multiple correlation *(CMC)*; and (iv) root mean square error (RMSE). Secondary outcome measure was running economy. Outcome measures were analysed with an independent samples t-test, with group allocation as a fixed factor. Standardised mean differences (SMD) were calculated using pooled standard deviations.

RESULTS: Eight triathletes exhibited altered neuromotor control at baseline testing and were randomly allocated into control or plyometric groups. There were no significant differences between groups at baseline for all measures (Table 2). There were significant differences between groups at 8 weeks that favoured plyometrics for the primary neuromotor outcomes of mean EMG amplitude and RMSE ($p = 0.043$, SMD = 1.2 and $p = 0.008$, SMD = 2.9, respectively – see Table 2). Altered neuromotor control was corrected at 8 weeks in 100% triathletes in the plyometric group, compared to 40% in the control group. Running economy was not significantly different between groups at 8 weeks ($p \geq 0.17$).

Table 2. Mean (SD) pre and post data for the control and plyometric groups and mean difference (95% confidence intervals) between groups for primary and secondary outcomes measures at 8 weeks.

	Mean (SD)		Mean (95% CI) difference
	Pre	Post	
Primary Outcome Measures			
Mean EMG amplitude (%)			
Control	-14.5 (3.7)	-9.2 (5.9)	-9.1 (-17.8 to -0.4)*
Plyometric	-14.3 (6.6)	-0.5 (1.5)	
CMC			
Control	0.64 (0.15)	0.81 (0.08)	-0.11 (-0.2 to 0.01)
Plyometric	0.64 (0.18)	0.92 (0.03)	
RMSE (%)			
Control	11.7 (3.3)	8.1 (1.1)	3.0 (1.1 to 4.9)*
Plyometric	12.4 (3.1)	5.1 (0.9)	
Secondary Outcome Measure			
Running economy ($ml.min^{-1}.kg^{-1}$)			
Control	-0.2 (1.7)	1.24 (1.6)	1.8 (-1.0 to 4.5)
Plyometric	-0.64 (1.6)	-0.53 (1.4)	

DISCUSSION: The addition of plyometric exercises to regular endurance training corrected the presence of altered neuromotor control when running off-the-bike. Following the intervention period, all triathletes in the plyometric group exhibited muscle recruitment patterns during running after cycling that closely resembled the recruitment patterns used during an isolated run. Despite the improvements in neuromotor control, 8 weeks of plyometric training did not enhance or impede running economy off-the-bike.

CONCLUSION: Combined plyometric and endurance training corrected neuromotor control in those triathletes in which it was previously aberrant. The favourable neuromotor outcomes did not translate into improved or impaired running economy, but may have utility in the prevention of injury and/or rehabilitation.

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Running

New Investigator Award

TRAINING FOR THE BIKE TO RUN TRANSITION IN TRIATHLON

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The purpose of this study was to examine the effect of a practice regimen that targets the bike-to-run transition for triathlons; known as brick workouts. The principle of specificity suggests that since this skill is a critical transition in a triathlon, having further impact on the subsequent running section, practicing this skill is vital for success. Moreover, the identification of performance parameters that quantify a successful transition will serve to maximize practice efficiency. Subjects (N=12) performed either brick workouts or single event training, to examine their effects on the bike-run transition. Our results indicate that the brick workouts had a positive effect by eliciting an increased adaptability in knee behavior in response to the transition from cycling to running. Quicker adoption of efficient running mechanics may ensue, leading to less fatigue and greater performance.

KEY WORDS: approximate entropy, transition, brick, movement, training, triathlon.

INTRODUCTION:

Transitioning from biking to running during a triathlon can be a difficult endeavor for a novice, or even a professional triathlete. The difficulty in this transition is highlighted by the retention of the movement pattern characteristic of biking, well into the initiation of the running segment. At minimum, easing this transition can bring added comfort to the athlete, and at maximum it could mean the energetic difference between victory and defeat. As known in the triathlete community as the “brick” workout, this type of transition training focuses on the repeated alternation of bike and run training during a single practice session. This is in stark contrast to the traditional training approach of single skill training during separate practice sessions, where the transition phase receives little attention. Brick training brings specific practice efforts to the optimization of the transition phase, in order to alleviate the carry-over detrimental effects of distance cycling on the subsequent running performance. However, even though brick training is developing wide spread use in the triathlon community, and is strongly purported by the principle of specificity, no definitive research could be found to support the use of brick workouts and its effects on running performance following the bike-run transition. Thus, the purpose of this research was to document the effectiveness of brick workouts, while concurrently presenting a novel measure of performance which could continue to benefit the analysis of triathletes' performance.

METHOD:

Participants and Grouping:

Seven male and five female college-aged triathletes were split randomly into two groups (experimental brick training group & control training group). Maximal oxygen uptake (VO_{2Max}) was recorded pre-post training in order to ensure standardized fitness between groups.

Procedures: Triathletes in both groups engaged in 6 weeks of triathlon preparation using one of two training methods; brick or traditional. Relative practice intensity was maintained between the groups by providing similar training volumes during specific practice (2 hours per week), documentation of miles and minutes of training outside of the specific practice (both cycling and running), and the instructed regulation of 80% heart rate max during specific practice.

The brick training group practiced the run-bike transition by performing transition training, alternating 10 minute bike with 10 minute run, for one hour twice each week. The traditional training group practiced bike and run workouts for one hour each week, on separate days. SIMI motion analysis was used to record lower extremity kinematics (at 60 Hz) during each triathlete's initial running performance after the bike-run transition, both pre and post 6 weeks of training. For both tests, participants cycled for 40 minutes at 90% max heart rate (to simulate true race conditions), and were then filmed for the first minute of the following run. To aid in digitizing, retroreflective markers were fixed to known body landmarks of the lower extremity (greater trochanter, lateral epichondyle, and lateral maleolus). SIMI was also used for manual digitization and calculation of knee angle; used for subsequent analysis.

Design: The current investigation utilized a 2 X 2 design (time X training), with a corresponding ANOVA used to test for differences across three dependant variables. Time (the within subjects factor) had two levels, pre and post training. Training type (the between subjects factor) also had two levels, traditional and brick training. Three dependent variables were calculated from the knee angle data collected during the first minute of running following the transistion phase from a 40 minute bout of cycling. Dependent variables included 1) knee angle range (max – min knee angle position), 2) knee angle coefficient of variability (standard deviation of position/mean position), and 3) approximate entropy (ApEn; a nonlinear measure of knee angle temporal predictability, i.e. consistency of position during recurrent cycles).

Approximate Entropy (ApEn)**:** This value characterizes the complexity of variability within the signal; the degree to which a signal remains self-similar through time (Vaillancourt, Newell, 2000). Higher values indicate that the behavior is irregular and lower values indicate maintained consistency. ApEn is computed using algorithms by Pincus (1994), implemented in MATLAB. Stergiou (2004) report that ApEn is useful for characterizing the health and optimal performance of a biological system including heart rhythm, standing posture, and gait.

ApEn has been used to examine heart rate complexity proceeding atrial fibrillation showing that a period of very low ApEn precedes atrial fibrillation (Tuzcu, 2006). Studies of postural control have found that after cerebral concussion an athlete's center of pressure oscillations, measured by ApEn, are significantly more rigid up to 96 hours post-injury, even when the athlete appears steady (Cavanaugh, 2006).

Analysis: Linear analysis of knee angle was performed by calculating the average of both range and coefficient of variability for each group during the first minute of running after 40 minutes of cycling, both pre and post training. Non-linear analysis was calculated across the whole of the first minute of running after the transition phase.

RESULTS:

No overall group changes in VO_{2Max} were observed. In addition, exercise outside of the training protocol did not vary between groups. Linear analysis of knee angle range and coefficient of variability showed no measureable difference for the two conditions within groups (Figures 1 & 2), indicating that the knee moved within the same volume of space regardless of the training. However, the non-linear measurement of ApEn (Figure 3) showed a difference between pre and post training within the brick training group, t (5) = -3.49, p=0.017, but not for the traditional training group, t (5) = 0.49, p=0.640. The change in ApEn value indicated that the brick training group experienced changes to the temporal organization of knee movement, decreased rigidity, as a result of training. Although pre-test values exhibit differences between the two groups, the study participants were quite similar in terms of age, experience, and anthropometrics and were placed randomly into the groups. Moreover, the goal of this study is to focus on changes in performance due to training type. Further investigation and analysis may be needed to uncover the source of an athlete's specific *a priori* mechanical approach to running.

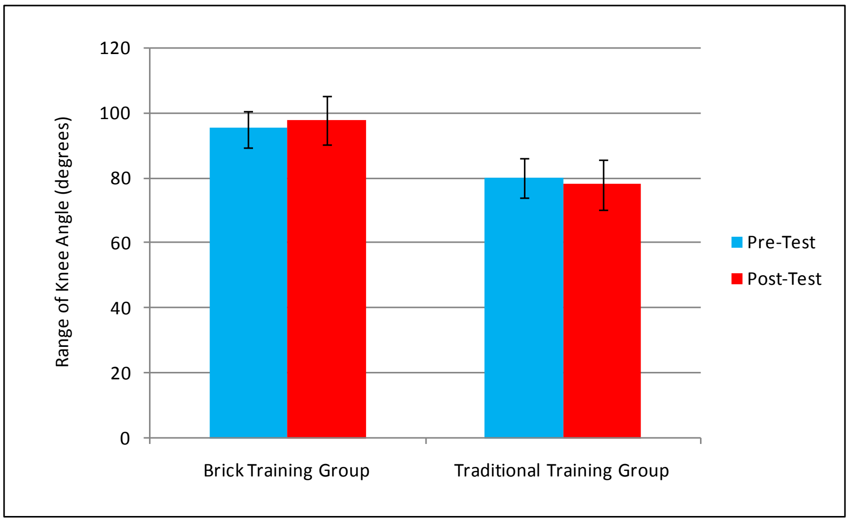

Figure 1: Knee Angle Range

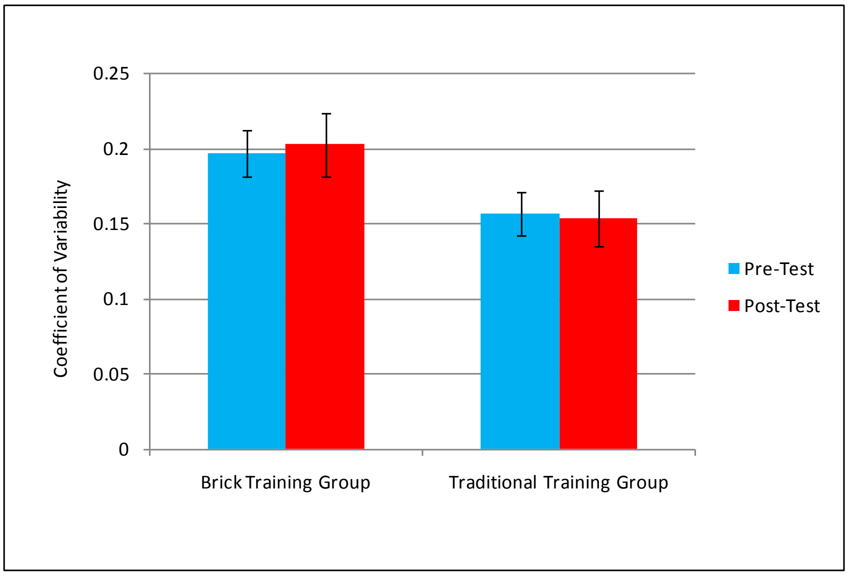

Figure 2: Knee Angle Coefficient of Variability

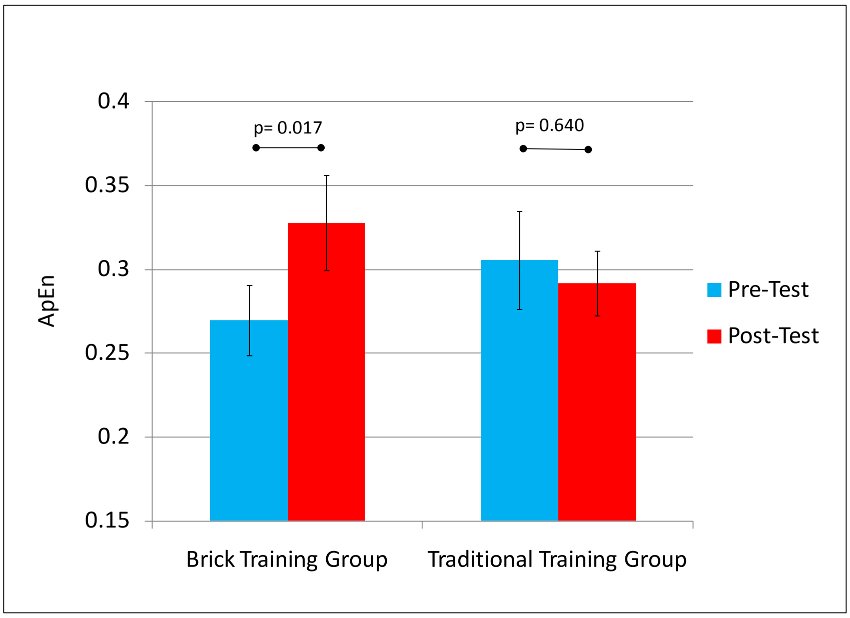

Figure 3: Knee Angle ApEn

DISCUSSION: Approximate entropy characterizes the regularity, or predictability, of a behavior. Lower values of ApEn indicate a system that is highly recurrent, repeating nearly the identical behavior throughout subsequent iterations. What this means in terms of running mechanics, is that lower ApEn values indicate rigid, non-adaptive strategies. Specifically in the context of the bike-run transition phase of the triathlon, this rigidity indicates a reduced ability of the neuromuscular system to transition from cycling mechanics to running mechanics, due to being "stuck" in the repeated engagement of cycling behavior. Slowed, inefficient transitions between these performance mechanics cause drastic deficits to performance, serving as a source of pain and discomfort and increasing propensity for injury. Additionally, the perseverance of cycling mechanics into the running section is quite expensive, leading to large consumptions of oxygen and energy that could instead be used to heighten the athlete's performance.
The increased values of ApEn found following brick training suggest that individuals in this group have acquired a new, transition-specific flexibility in knee angle behavior. Reduced rigidity following the cycling section of the race will allow for a quicker, smoother, and more efficient transition into steady state running mechanics. Coupled with the observation of unchanged range and coefficient of variation, these results suggest that it is the temporal organization of knee joint behavior which differs after training, with the opportunity to influence performance.

CONCLUSION: The decreased rigidity in knee movement behavior found after training for the transition through brick workouts reflects an increase in fluidity and flexibility of movement. Increasing knee angle ApEn indicates greater adaptability during the run after the transition, indicating the potential for quicker, more efficient transitions into steady state running mechanics. It is important to mention that linear analysis was unable to detect these subtle changes that occur resulting from brick training. Non-linear analysis may detect this information readily and help to explain why so many triathletes participate in brick training. Upon further investigation, ApEn may become a critical evaluative measure of performance for triathletes.

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EFFECT OF RESPIRATION DYNAMICS ON POSTURAL CONTROL FOLLOWING A 5K RUN

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KEYWORDS: Postural Control, Fatigue, Nonlinear Dynamics

INTRODUCTION: Research has shown postural control during upright stance can be diminished for up to twenty minutes following aerobic exercise of different types, intensities, and durations (Lepers et al., 1997; Nagy et al., 2002). Researchers have posited that this is caused by neuromuscular changes associated with aerobic exercise and fatigue such as the reduced excitability and central drive to peripheral muscles (Lepers et al., 2002), vestibular desensitization (Lepers et al., 1997), and peripheral somatosensory desensitization (Lepers et al., 1997). However, no research has measured or attempted to control for the influence that changes in respiration dynamics (e.g., rate and volume) alone might have on postural sway. The aim of the current study was to examine these effects in order decipher whether changes to postural control following intense aerobic exercise (a 5-kilometer run performed with maximal effort) can be attributed to effects of exercise and fatigue or simply changes in respiration.

METHODS: Eighteen (M=9, F=9) healthy college-aged students underwent two experimental sessions. In the first session participants completed a 5k treadmill run (average run time = 28:00 min). Center of Pressure (COP) and respiration rate/volume were recorded prior to the run (resting), immediately after completing the run (0-min.) and at 2, 5, and 10 minutes post-run while participants stood bipedal on a forceplate for trials of 30 seconds (COP data sampled at 100Hz). In a follow-up session (one week after the initial session) participants returned and were asked to replicate the exercise-induced respiration dynamics obtained during the first session while COP was recorded for trials of similar durations. A metronome was used to help participants replicate their respiratory rates and feedback was provided by experimenters to aid in replicating respiratory volume. Anterior-Posterior (AP) and Medial-Lateral(ML) COP data were analyzed with Sample Entropy (SampEn – a measure of sway complexity) (Richman & Moornman, 2000), Root Mean Square Amplitude (RMS – a measure of average sway ‘area’), and Path Length (a measure of overall ‘amount’ of sway). Inferential analyses were conducted using a set of 2(Condition; 5k run vs. Replicated) x 5 (Time; resting, 0, 2, 5, and 10-min) repeated measures ANOVAs for each dependent variable followed by a set of planned contrasts for all Condition and Time pairings (α=0.02).

RESULTS & DISCUSSION: Following the 5k run both respiration rate and volume were increased compared to resting levels up to 10-min. post run. No significant differences in respiration rate and volume were found between the first and follow-up sessions for any time period. This later finding indicates that participants successfully replicated their respiration dynamics in the follow-up session. Results from analysis of COP data revealed that increases in respiration rate and volume alone caused changes to postural sway. The measures and time periods where changes occurred are indicated in Figure 1 (*).These changes included increases in the amount (increased Path Length), but decreases in the complexity (decreased SampEn) of sway, both of which can be interpreted to reflect decreased postural control. However, results showed that changes in respiration dynamics alone could not account for all the changes that occur to postural sway following intense aerobic exercise. Instead, following the 5k run participants showed further decreases in sway complexity, and increases in the ‘amount’ and ‘area’ of sway, as see in Figure 1 (^).

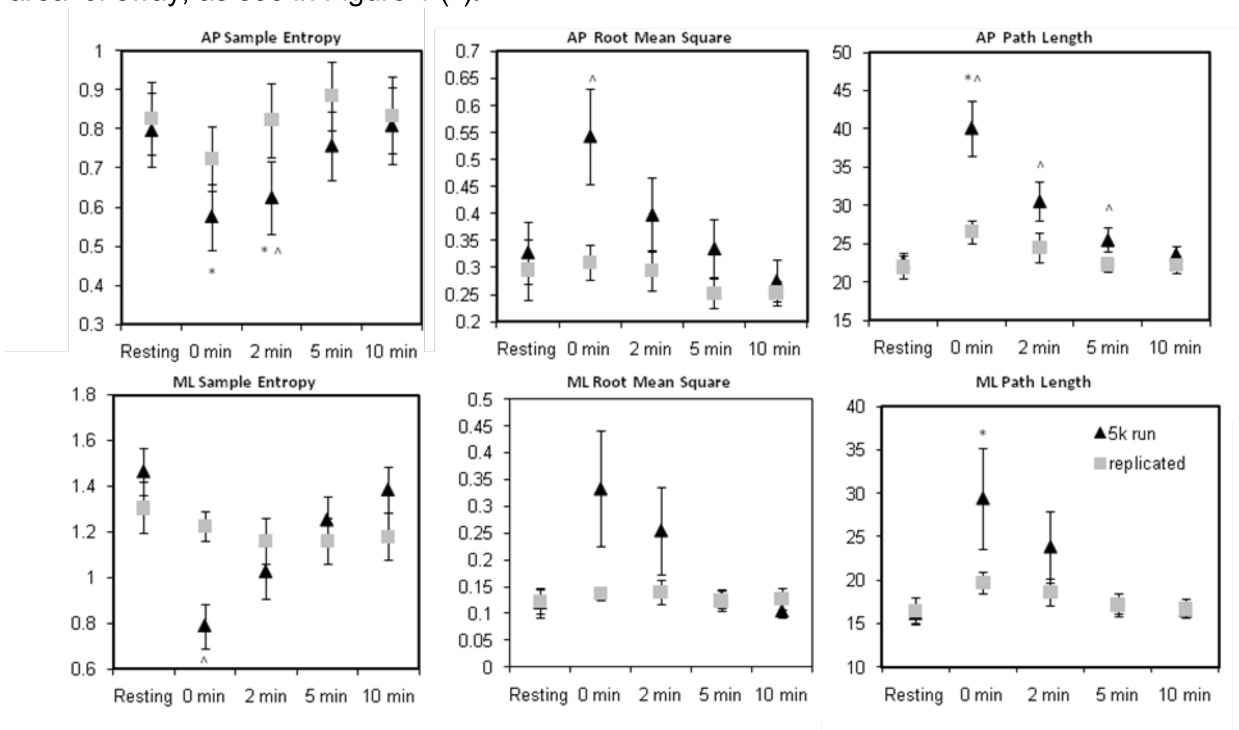

Figure 1. Mean (±SE) for each COP postural sway measure for 5k-run and Replicated conditions.
* indicates significance ($p \leq 0.02$) between Replicated and Resting values.
^ indicates significance ($p \leq 0.02$) between Replicated and 5k-run values.

CONCLUSION: These findings indicate that postural control may be compromised following intense aerobic exercise for a brief period of time (less than 5-min), but that these effects cannot be solely accounted for by changes in respiration leaving open the possibility that other neuromuscular changes associated with exercise and fatigue (e.g., vestibular desensitization) may negatively affect postural control. Although more work is needed to reveal exactly what these changes are and how they specifically effect postural control. This study may be of practical importance for those who engage in tactical precision activities following long or intense bouts of aerobic exercise (e.g., target shooting for biathletes, law enforcement agents, military personnel engaged, etc.) since our results show that postural control may be diminished by aerobic exercise in ways above and beyond that which can be attributed solely to respiration.

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THE INTER-DAY RELIABILITY OF A METHOD USED TO DETERMINE VERTICAL, KNEE AND ANKLE STIFFNESS DURING RUNNING

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The purpose of this study was to investigate the inter-day reliability of common methods used to measure vertical (K_{vert}), ankle ($K_{AnkBrak}$) and knee joint ($K_{kneeBrak}$) musculoskeletal stiffness during running. Currently there is a paucity of research investigating lower extremity musculoskeletal stiffness during running tasks moreover, the reliability of the methods and findings has not been adequately established. Twenty seven active male participants performed ten trials of running at a velocity of 3.35 m/s during two sessions, separated by 3-7 days. Five trials were completed with the one foot contacting the force plate embedded in the laboratory floor, with the remaining trials to test the other foot contact. Excellent reliability was shown for K_{Vert}, however $K_{AnkBrak}$ and $K_{KneeBrak}$ exhibited average and poor ICC's.

KEYWORDS: vertical stiffness, joint stiffness, reliability, running.

INTRODUCTION: Lower extremity musculoskeletal stiffness (MSS) is considered to be an important factor in musculoskeletal performance and possibly injury (Butler, Crowell, & McClay-Davis, 2003). During activities such as running there is an energy exchange between the muscles, tendons, ligaments, and bones in order to provide efficient movement (Cavagna & Kaneko, 1977). In this context, the body is modelled as a mass supported by a spring which is modelled as the leg. Vertical stiffness (K_{Vert}) is then used to describe the linear motions of the body that occur only in the vertical direction. K_{vert} is the ratio of peak vertical ground reaction force and centre of mass (COM) displacement. During running, K_{Vert} has been shown to increase with increasing running velocity (Cavagna, Heglund, & Willems, 2005). K_{Vert} has also been shown to increase with the physical demands of the activity (Kuitunen, Komi, & Kyrolainen, 2002), with running economy (Dutto & Smith, 2002) and with increasing stride length (Farley & Gonzalez, 1996).
Joint stiffness is used to model the individual joints as rotational springs (Arampatzis, Schade, Walsh, & Bruggemann, 2001b). Joint stiffness can be calculated by the ratio of joint moment and joint angular displacement. During running, knee joint stiffness has been shown to increase with faster running velocities whilst, ankle joint stiffness remains constant (Gunther & Blickhan, 2002). The change in knee joint stiffness is mainly due to the decreased knee flexion when compared to ankle flexion, thus suggesting that knee joint stiffness is the controlling factor of lower extremity (leg) stiffness during running.
Vertical and joint MSS are commonly investigated for running tasks. Whilst the reliability of specific biomechanical methods to measure ankle joint stiffness during hopping tasks has been established (McLachlan, Murphy, Watsford, & Rees, 2006), there appears to be a paucity of research investigating the reliability of methods assessing MSS during running. Therefore, the current study attempted to establish the reliability of a laboratory-based biomechanical assessment of vertical, knee and ankle joint stiffness.

METHOD: Twenty seven active males from various sporting backgrounds were recruited from the Australian Catholic University School of Exercise Science to participate in this study (age: 22.3 ± 3 years, mass: 74.7 ± 5.6 kg, stature: 1.79 ± 0.7 m). The sample size for this study was based on sizes used by similar studies investigating the reliability of methods assessing MSS (McLachlan et al., 2006). The study was approved by University Ethics Committee and all participants provided written informed consent prior to participating in this

study. All participants were injury-free at the time of testing and had not missed a training session or game in their respective sports six weeks preceding the time of testing.
The participants were required to attend two days of testing at the same time of day. The time between testing sessions was 3 – 7 days to avoid any effects from the previous session. A standardized warm-up was performed consisting of 5 minutes of cycling on an ergometer (Monarch AB, Sweden) and 5 minutes of stretching prior to testing. The participants were required to run at 3.35 m/s (Williams, McClay Davis, Scholz, Hamill, & Buchanan, 2004) along a 10 m runway. This speed was set due to the influence of running velocity on vertical, leg and joint stiffness (Brughelli & Cronin, 2008). Running speed was determined by two sets of timing gates (Smart Speed, 1.8 MHz, Fusion Sport, Australia) which were placed 5m apart. The force platform (Kistler, model 9268BA, Switzerland) was positioned exactly in the middle of the two timing gates. The participants were instructed to accelerate to reach the required speed and pass through the gates at that constant speed. The trials were accepted if they were within ± 5% of the target speed (Williams et al., 2004). As many practice trials as necessary were given for the participant to become familiar with the protocol. Ten trials were performed in total with the participant contacting the force plate with the left foot (five trials) and then the right foot (five trials). The order of which leg was recorded first was randomised to reduce the possibility of an order effect. Any trials that exhibited a forefoot ground contact pattern were excluded from the study due to the potential influence of a differing foot strike pattern on running kinetics and kinematics (Gunther & Blickhan, 2002).
A six-camera VICON motion analysis system (Oxford Metrics Limited, U.K.) was used to collect kinematic data at 100 Hz. Retroreflective markers were placed unilaterally on each segment of the lower body according to the requirements of the plug-in-gait model of VICON. The participants were instructed to run and look ahead to avoid targeting the force plate, and only trials that exhibited a clean foot contact were included. The force data was sampled at 1000Hz and contact with the force plate was set using at 15N trigger.
The kinetic and kinematic data were processed using custom written software (LabVIEW, National Instruments, Version 8.2, U.S.). Force data were filtered using a low-pass filter at 50Hz (Williams et al., 2004). The spring-mass model was used to represent the overall stiffness of the leg (Butler et al., 2003). Leg stiffness was calculated by

$$K_{Vert} = \frac{F_{peak}}{\Delta L}$$

where F_{peak} is the peak vertical ground reaction force, and ΔL is the displacement of the centre of mass during the braking phase of ground contact. The displacement of the centre of mass was calculated by integrating the vertical acceleration twice with respect to time (McMahon & Cheng, 1990). The braking phase was defined as from initial ground contact to maximum joint flexion. Joint stiffness was calculated as the ratio of joint moment (ΔM) to angular displacement ($\Delta\theta$) as shown in the equation below (Gunther & Blickhan, 2002).

$$K_{joint} = \frac{\Delta M}{\Delta \theta}$$

Data from the left and right legs were then averaged prior to the statistical analyses.
All statistical data were analysed using SPSS (Version 17.0, Chicago, IL). The data was checked for normality using the critical appraisal approach recommended by Peat and Barton (2005) and inter-day reliability was evaluated using intra-class correlations (ICC). The effect size (ES) was also calculated to determine if any trends existed and the magnitude of the trend. The coefficient of variation (CV_{ME}%) was also calculated to reveal the measurement error. Measurement error calculation (ME) is outlined in Peat and Barton. All statistical significance was set at $p < 0.05$.

RESULTS: This study investigated the reliability of musculoskeletal stiffness measures commonly reported during over-ground running involving single trials. Descriptive and reliability statistics for K_{Vert}, $K_{AnkBrak}$ and $K_{KneeBrak}$ over the two testing sessions are shown in Table 1.

Table 1. Musculoskeletal stiffness reliability during over-ground running for physically active young adult males.

	Day 1		Day 2					
Test	Mean	SD	Mean	SD	ICC	ME	CV_{ME}%	ES
K_{Vert} (kN/m^{-1})	26.82	2.10	26.80	2.50	0.88	1.01	3.77	-0.01
$K_{AnkBrak}$ (Nm/deg^{-1})	11.89	7.59	10.33	4.68	0.30	6.45	58.02	-0.26
$K_{KneeBrak}$ (Nm/deg^{-1})	13.01	5.16	10.78	4.07	0.28	4.25	35.71	-0.48

The K_{Vert} values between the two sessions showed little variation whilst joint stiffness values for the ankle ($K_{AnkleBrak}$) and knee ($K_{KneeBrak}$) were 13.12% and 17.14% less during session two when compared to session one. K_{Vert} reliability results indicated that the method was reliable with excellent ICC results (r = 0.88), low effect size (ES, -0.01) and small percentage variation between trials (CV_{ME}%, 3.77%). Results for joint stiffness however, revealed unreliable figures. $K_{AnkleBrak}$ returned ICC results that reflected poor reliability (r = 0.28) and a large percentage of variation between trials (35.71%). $K_{KneeBrak}$ returned an average ICC result (r = 0.30), medium ES (-0.26) and a larger percentage variation between trials (58.02%).

DISCUSSION: This study attempted to investigate and establish a reliable testing protocol for measuring lower extremity musculoskeletal stiffness during over-ground running at 3.35 m/s. Reliability statistics for K_{Vert} suggests that the methodology used in this study has very good repeatability. Alternately, K_{Knee} and K_{Ankle} reflect poor inter-session reliability. This poor repeatability may be a result of the large variations in CV_{ME}% and ES between test one and two for K_{Knee} and K_{Ankle}. This may suggest that a separate familiarisation session is necessary to improve the variation between tests.

K_{Vert} has been revealed to range from approximately 10 – 40 kN/m^{-1} at similar running velocities as the present (Arampatzis, Bruggemann, & Metzler, 1999; Divert, Baur, Mornieux, Mayer, & Belli, 2005; Morin, Dalleau, Kyrolainen, Jeannin, & Belli, 2005). This research compares well to the average of 26.81 kN/m^{-1} identified in this study. When comparing the current findings of both K_{Knee} and K_{Ank} levels to previous research the results are mixed, however. The averaged $K_{KneeBrak}$ level of 11.90 kN/m^{-1} compares well with 13.0 and 10.0 kN/m^{-1} shown in past research (Arampatzis et al., ; Gunther & Blickhan). Results for $K_{AnkleBrak}$ (11.11 kN/m^{-1}) are slightly larger or smaller than those previously reported (7.3 and 15 Nm/deg^{-1})(Arampatzis et al., ; Gunther & Blickhan). Discrepancies in these figures when compared to other research should be considered in light of alternate methodologies and participant cohort. For example, the study by Gunther and Blickhan involved a smaller cohort of 12 participants whereby four females were included. The study by Arampatzis and colleagues also included a population of 13 trained male runners which are again different to the cohort investigated here. Furthermore, some methodological differences may also explain the differences in MSS levels in this study as it was the only to use three-dimensional infrared motion analysis equipment to concurrently measure of K_{Vert}, $K_{AnkBrak}$ and $K_{KneeBrak}$.

CONCLUSION: The current study has concluded that the method detailed in this paper is reliable in measuring vertical leg stiffness during running in healthy males. Adopting the strictly controlled methodology outlined above provides the practical application of a very reliable K_{leg} assessment for over-ground running at 3.35 m/s. However, the method is deemed unreliable for determining knee and ankle stiffness. This is particularly important for future research interested in using an intervention whereby the expected MSS changes are smaller than the difference between sessions for ankle and knee stiffness. Caution must also be shown when interpreting the current results as the findings may only be applicable to research involving a similar testing protocol or population. Future research considerations might include the assessment of alternate experiment protocols for measuring ankle and knee stiffness or the inclusion of familiarisation session to reduce trial-to-trial variability.

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THE TRUNK ORIENTATION DURING SPRINT START ESTIMATED USING A SINGLE INERTIAL SENSOR

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KEY WORDS: trunk orientation, sprint start, block acceleration, inertial sensor

INTRODUCTION: Sprint start and block acceleration are two very important phases which could determine the result of a sprint. Tellez & Doolittle (1984) showed that these two phases account for 64% of the total result for a 100m sprint. Sprinters have to move from a crouch to a standing position, trying to reach their maximal velocity as fast as possible. Many authors have delved into the biomechanical factors concerning both phases (Fortier et al., 2005; Harland & Steele, 1997; Schot & Knutzen, 1992). Trunk orientation is considered by coaches one of the key elements in moving from the crouch to the upright position, however only a few studies focused specifically on this parameter (Čoh et al., 1998; Čoh et al., 2006; Natta et al., 2006). Moreover, the experimental setups used in the latter studies are quite cumbersome and limited in terms of acquisition volume (motion capture systems, high-speed cameras or optical contact time meters), therefore, they are hardly usable during everyday training sessions. Wearable inertial measurement units (IMU), that embed 3D linear acceleration and angular rate sensors (accelerometers and gyroscopes), can be effectively used to perform in-field biomechanical analysis of sprint running, providing information useful for performance optimisation and injury prevention. In particular, IMUs provide an estimate of body segment rotations relative to an inertia system of reference with one axis oriented as the gravitational field. The aim of this pilot study is to validate the use of a single IMU to estimate the trunk orientation angle in the progression plane during a sprint start from the blocks.

METHOD: A female subject (age=29yrs, m=56kg, h=1.71m) performed four in-lab sprint starts from regular starting blocks. The lab floor was covered with a felt carpet and the blocks were fixed directly into the floor. The two main positions of the sprint start ("on your marks", OYM and "set", SET positions) and the first three steps of each start were analysed. An IMU (Freesense, Sensorize, Italy) was positioned on the lower back trunk (T10 level). In order to limit the sensor movement relative to the underlying bone, various fixing methods were tested. Finally, a memory foam material was placed between the trunk and the sensor, which was then fixed with an elastic belt. The validation of the IMU estimates was performed by means of four retro reflective markers placed on the sensor. Their movements were tracked using a stereophotogrammetric system (Vicon MX3, Oxford, UK) considered as the reference, and the sensor global orientation was then calculated. The IMU orientation was supposed to provide information about the trunk orientation under the assumption of trunk rigidity. The movement was then analysed in the sagittal plane, as this is supposed to be aligned with the average plane of progression. After having identified the static and dynamic phases in each trial, acceleration and angular velocity measures provided by the sensor were used to identify trunk orientation as follows. When the sensor inertial acceleration was close to zero, i.e. during the sprint start static phases (OYM and SET positions), the accelerometer measured the inclination of the sensor relative to a vertical line defined according to gravity, thereafter called β. A quaternion based algorithm (Favre et al., 2006) was implemented in order to compute β (usually referred to as pitch). Conversely, when the sensor underwent a motion that generated inertial accelerations, i.e. during the dynamic phases (transitions between OYM and SET positions, and blocks acceleration) the trunk orientation angle β was estimated integrating the angular velocity signal provided by the gyroscopes. This allowed for the reduction of the integration interval and, thus, for a decrease of drift errors associated with the numerical integration process. In order to test the accuracy of the estimated angles during the static phases, an average value of the pitch angle β during the OYM and SET positions was considered: β_{OYM} and β_{SET}, respectively. The average of the absolute difference between the reference and the IMU estimates (e=|Vicon-IMU|), referred to as error (e), was calculated both for β_{OYM} and β_{SET}. To assess the curves similarly, the Root Mean Square Error (RMSE) between the reference and the estimated angles was computed.

Moreover, in order to take the curve temporal shift into account, the Pearson's product-moment correlation coefficient (r^2) was also calculated.

RESULTS: The Pearson's correlation coefficient and the Root Mean Square Error (mean and standard deviation) between reference and IMU angle estimates were respectively: r^2 = 0.992±0.006 and RMSE = 0.149±0.051deg. Trunk orientation angle curves for one trial, obtained from the reference measurement (solid line) and the IMU (dashed line) are shown in Figure 1. The pitch angle was considered to be zero when the sensor was in a horizontal position; clockwise rotations correspond to positive angles. Reference and sensor estimates, and absolute error (e) for β_{OYM} and β_{SET} are reported in Table 1 (mean and standard deviation).

Fig.1 – Flexion-extension angle

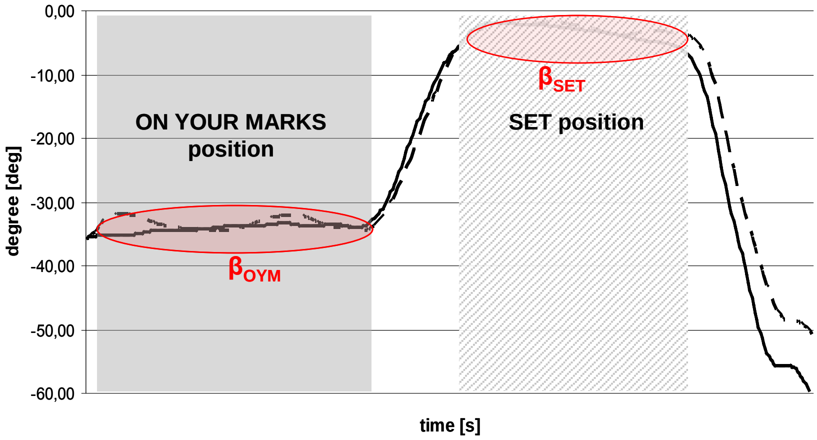

Table 1 – Absolute error between reference and IMU estimates for β_{OYM} and β_{SET}

	reference	IMU	error
β_{OYM} [deg]	-35,5±4,7	-33,8±3,4	1,1±2,1
β_{SET} [deg]	-1,1±3,2	-2,0±4,5	1,2±1,6

DISCUSSION: Average absolute errors during the static phases of the start were less than 2 deg, meaning that sensor inclination estimated from acceleration signals was accurate. The accuracy of the trunk orientation angle during the dynamic phases depends mainly on the on the integration interval duration and unit fixing. The tested memory foam material resulted in a good solution to limit the noise introduced by soft tissue oscillations. Moreover, since the subject went from the "set" to the upright position in less than three seconds, the drift errors typical of the numerical integration process were limited. This is particularly true for professional sprinters.

CONCLUSIONS: The study shows that it is possible to accurately estimate the trunk orientation angle during a sprint start from the blocks by using a single IMU. However, the effect of the assumption of trunk rigidity on errors remains to be verified. Future works will concern the validation of the method in-field and the analysis of the trunk orientation influence on performance, with the aim to use the sensor with professional sprinters during all-out sprint starts.

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TOWARDS AN AUTOMATED FEEDBACK COACHING SUPPORT SYSTEM FOR SPRINT PERFORMANCE MONITORING

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The purpose of this study was to investigate the feasibility of developing a cost-effective, automated performance feedback system to support sprint coaching. The proposed system is designed to deliver step length, step frequency, contact time and 10 m split time information of multiple athletes training on an indoor track. An integrated systems approach was chosen combining the novel Pisa Light-Gate (PLG) and Step Information Monitoring Systems (SIMS). Current results indicate data accuracy of RMS 1.662 cm for step length, RMS 0.977 ms for foot contact time and a split time detection accuracy of 8.45 ± 6.85 ms. These results suggest that the proposed integrated system, using off-the-shelf equipment, would go beyond currently available coaching tools by providing automated and highly accurate sprint performance information for multiple athletes.

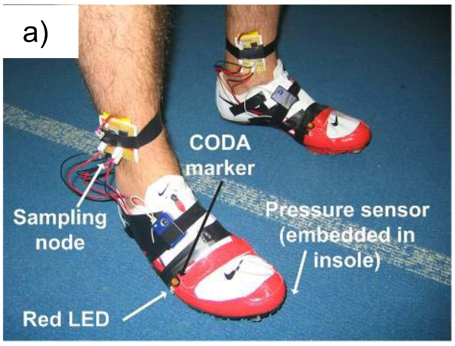
KEYWORDS: foot contact time, split time, step length, step frequency, athlete support.

INTRODUCTION: Performance monitoring systems for athletics have, to this point in time, provided either very basic information, involved considerable time to set up or required extensive post-processing efforts and a high cost of purchase. Thomson et al. (2009) showed that elite sprints coaches are keenly interested in a number of performance and technique related parameters in order to quantitatively monitor the development of their athletes over time. However, the restrictions of currently available systems in terms of their ease of use and databasing capabilities have largely prevented their routine use in training. Furthermore, thus far the information that can be obtained by the coach is largely restricted to timing data over specified distances and no quantitative feedback information relating to aspects of technique have been made accessible to the coach.

One aspect of sprinting technique that has been highlighted as being of interest to the coach is the interaction of the foot with the ground (Thomson et al., 2009). The interaction of the foot with the ground is represented by a number of variables including step length (SL), step frequency (SF) and contact time (CT). These parameters and their interaction for the sprint start and maximum velocity phases have received substantial interest in the literature (Ae et al., 1991; Bezodis et al., 2008; Coh et al., 2006; Cronin and Hansen, 2006; Frishberg, 1983; Hunter et al., 2004; Kristensen, 2006; Kuntze et al., 2009; Luhtanen and Komi, 1978; Mann and Herman, 1985; Salo and Bezodis, 2004). However, due to a lack of long-term, inclusive set of sprint performance data, there is as yet no clear consensus as to the benefit and feasibility of altering any one of these parameters in order to enhance athlete sprinting velocity.

In order to address the need for enhanced accurate information availability for sprint coaching using cost-effective equipment, it was decided to utilise an integrated systems approach combining a permanently installed multi-lane split-time monitoring system, the Pisa Light Gate (PLG) system (Cheng et al., 2010), with a novel foot parameter detection system, the Step Information Monitoring System (SIMS). PLG supports long term databasing and on-line review of 10 m split time measurements for a 60 m indoor running straight for multiple athletes running at a time. The system is operated using a WiFi controller (iPod Touch, Apple Inc.) which also allows for the immediate display of timing information. The SIMS system uses integrated track-side low-cost cameras and on-body foot pressure sensors for detecting foot CT, SL and SF. The use of an integrated approach for the development of SIMS allows for the utilisation of the existing PLG hardware and software infrastructure providing an ideal basis for the expansion of sprint related measurement capabilities and the instantaneous display of feedback information. However, the suitability of cost-effective equipment for high performance sprint monitoring – which requires high level of accuracy - is not known. The aim of this study was to use a prototype of the proposed SIMS system to assess the feasibility and accuracy of an integrated system capable of providing instantaneous performance and technique related feedback to the athletics coach.

METHOD: One recreational athlete (29 yrs, 81.0 kg, 1.84 m) gave written informed consent to participate. He was asked to perform six sprint start and maximum velocity runs on the 60 m indoor running track of the National Indoor Athletic Centre (NIAC) in Cardiff, UK. The participant performed his normal warm up routine before performing the sprint runs. Upon completion of the warm up the participant was fitted with two custom built pressure sensing insoles utilising force sensing resistors (FSR). These acted as controllers for two pairs of super-bright light emitting diodes (LEDs) which were mounted to the sprinting spikes worn by the athlete (one on the lateral side of right foot and one on the medial side of the left foot) triggering the activity of the LEDs during the support phase of the sprint cycle (see Figure 1a).
SL data were collected using SIMS, installed at the final quarter section (45 to 60 m) of PLG. SIMS consisted of a 15 Hz Point Grey video camera (CMLN-13S2C-CS) operated by a laptop computer. The camera was located in the roof section of the athletics centre giving a total field of view (FoV) of ~15 m. For FoV calibration, lane 4 of the running straight was marked at 0.5 m intervals using the lane edges and the centre of the lane (see Figure 1b). The lane was therefore divided into multiple virtual zones, enabling accurate camera calibration for SL calculation using a custom-designed, geometry-based calibration algorithm. SL was determined based on the LED light impulse during support: each red dot in the video images therefore represents the position on the ground where the foot has landed. SL was calculated based on the horizontal distance between two subsequent red dots in relation to the underlying calibration reference points (see Figure 1b), which are needed once when setting up the (fixed-location) camera.
SL validation data were collected using an automated 3D motion analysis system (CODA), recording the horizontal displacement of active markers (at 200 Hz) placed superiorly to the LEDs used for the SIMS system. Four CODA scanners were placed unilaterally along the side of the sprint straight providing a total FoV of ~16 m (see Figure 1b).



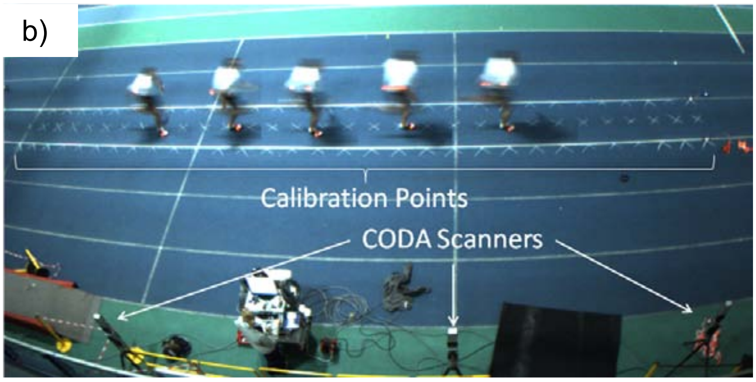


Figure 1. a) The SIMS insole-based pressure sensing prototype (with LEDs); b) a birds eye view of the total SIMS FoV showing the participant, calibration points and CODA scanner positions.

RESULTS: SL values computed from SIMS were compared to those from CODA (see Table 1). The absolute RMS error was larger for the entire 15 m FoV than the central 10 m section of the FoV due to the effects of lens distortion. Figure 2 displays all SL data for the central 10 m section of the FoV using SIMS and CODA.

Table 1. Difference in SL determined using SIMS and CODA

	15m FoV (cm)	10m FoV (cm)
RMS	2.417	1.662
Max	9.150	4.528
Min	0.001	0.001

DISCUSSION: The findings of this investigation indicate very encouraging initial levels of accuracy for the automated determination of SL using the SIMS system. Step length data were in agreement with the validation measurement technology (CODA) showing a comparative absolute RMS error of < 2 cm for a 10 m FoV (see Table 1). Importantly, no foot/ground contact events were missed by the SIMS system indicating that a 15 Hz sampling rate is sufficient for the determination of SL in sprinting with contact times unlikely to be less than 90-100 ms in duration (Kuntze et al., 2009). The level of accuracy at this early stage of system development may be improved by the use of additional reference points in the calibration field. It provides clear indication of the effectiveness of the automated detection algorithm for providing gold-standard comparable step information.
Furthermore, this level of automated SL determination accuracy is encouraging for the purpose of inclusion of SIMS within the proposed integrated system framework. Previous validation and application testing has demonstrated high levels of accuracy for the automated determination of 10 m split times using PLG (8.45 ± 6.85 ms) (Cheng et al., 2010) and automated foot contact time determination using FSR based insole technologies (absolute RMS 0.977 ms; max error 1.98 ms; min error 0.001) (Kerwin, 2009; Kuntze et al., 2009; Taherian et al., 2010). Importantly these levels of accuracy are achieved using affordable, and readily available technologies (SIMS camera ~£200; PLG retro-reflective light gates ~£60 each; and FSR insole hardware ~£5 per pair) which is essential for providing cost effective performance monitoring solutions to coaches and athletes.
Future developments of the system are focused on the full implementation of SIMS as a completely integrated part of the PLG – SIMS Integrated System (IS) and the expansion of SIMS to cover the entire 60 m length to match the measurement capabilities of PLG. The integration of SIMS derived SL information with PLG and FSR insoles will be a significant advancement of measurement capabilities in athletics. It creates the possibility of simultaneously collecting long-term data on a variety of sprint performance parameters including 10 m split times, step length, contact time, flight time and step frequency as well as step by step changes in athlete velocity. It is anticipated that the creation and regular use of this integrated system will advance our understanding of the contributors to athletic performance by providing long-term, multi-variable, athlete-specific performance data. This may aid in the assessment of the benefits and feasibility of alterations to any one of these parameters in order to enhance athlete sprinting velocity.

CONCLUSION: This study demonstrates the feasibility of developing a novel, accurate, measurement system using cost-effective equipment for the assessment of performance and technique in sprinting and related athletic disciplines. The integrated PLG and SIMS systems provide a unique opportunity to gather long-term athlete record data which is anticipated to be highly useful for the coach in order to track performance changes and gauge the effectiveness of training methods.

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ISBS 2010

Oral Session 10

Soccer & Tennis

'BEND IT LIKE BECKHAM': BALL ROTATION IN THE CURVED FOOTBALL KICK

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The objectives of this study were to quantify the 3D angular velocity and spin axes of a curved versus straight kick for goal in football. A 12 camera 250 Hz 3D Vicon motion analysis system recorded 4 semi-professional soccer players, as they performed 5 straight (S) and 10 curved (C) kicks. While the velocity of the ball was similar for both kicks (~20 m.s^{-1}), spin rate was significantly different (S=22.6 rad.s^{-1}; C=36.4 rad.s^{-1}). While the level of spin for the straight kick was surprising, the elevation angle of its spin axis was significantly lower in the straight compared with curved kicks (S=30.4°; C=62.6°).

KEYWORDS: Aerodynamics, football, ball rotation

INTRODUCTION: Football is frequently termed the 'world game' because of its playing numbers and spectator appeal, with scoring goals the key to success at all levels of play. The curved (C) kick is one of the key skills used to score goals and as such is an important aspect of the game. For such an important skill, the C kick has received significantly less attention from researchers compared with other types of football kicks. Successful C kicks rely first on a high velocity, off-centre contact and on the aerodynamic properties of the spinning ball, thereafter. The C kick is generally performed at sub-maximal velocities (Wang & Griffin, 1997), allowing the kicker more control over the position of the foot at ball contact. The few studies that have examined kicking a football with spin have recorded ball velocities between 15.1 and 25.4 m.s^{-1} (Asai et al., 2002; Griffiths et al., 2005), values lower than those measured for maximal instep kicks, which have been reported between 28.0 and 32.6 m.s^{-1} (Asai et al., 2002; Nunome et al., 2006). The further the foot contact point lies from the mid-line of the ball, the smaller the resultant ball velocity will be, while the spin rate will logically be greater owing to the eccentric force applied to the ball. There is a paucity of data investigating spin rate for curved kicks, with a single study establishing that the spin rate varies between 25 and 59 rad.s^{-1}(Griffiths et al., 2005). Other flight characteristics of the spinning soccer ball, such as the nature of the spin axis, have been neglected in the literature. The aims of this study were to identify the ball's 3D rotational and flight characteristics in a C compared with straight (S) kick for goal in football.

METHOD*:* Four semi-professional right footed soccer players aged 21.3 ± 1.0 years, height 173.5 ± 4.0 cm and mass 69.5 ± 10.1 kg, who were expert exponents of the curved kick were recruited. Testing was performed in an indoor laboratory that opened onto a grassed area, where the target goal was placed. The capture volume was calibrated using Vicon's standard procedures with the reconstruction volume large enough to record at least 5 m of ball flight post impact. A conventional size 5, FIFA approved ball was fitted with 4 hemispheric foam retro-reflective markers, so as to limit their effect on the mass and aerodynamic properties of the ball. A 12-camera 250 Hz, Vicon motion analysis system (Oxford Metrics Inc., UK) was used to track these markers, while a digital video camera, sampling at 50 Hz was positioned behind the kicker to record the ball's lateral deviation in the frontal plane (Fig 1). A 1.2 x 1.5 m target area was positioned in the top right hand corner of the goal. Following a warm up, participants completed S and C kicks at a set distance of 20.12 m from the goal (Fig 1). During the S kick, participants were instructed to aim for the target and to kick the ball without spin, using a technique they would normally use to perform an instep drive pass for distance in a game situation. Five successful S kicks that impacted the target area were recorded in order to establish baseline data. In the C kick, a 2 m vertical pole was placed between the kicker and the goal to represent the position where the outermost player in a defensive wall would stand if a direct free-kick was to be taken from the same position. The players were then instructed to curve the ball around this pole, with their preferred right foot, aiming at the same target area of the goal. The order of the kicks was randomised to eliminate order effects and successful kicks were required to curve around the pole and impact the target area.

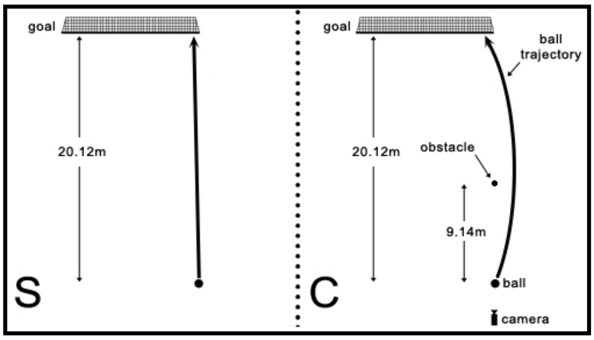

Figure 1. Experimental set-up of the motion capture laboratory for the straight and curved kicks.

A local coordinate system was established for the ball that enabled the angular velocity vector in the global coordinate system to be calculated using the methods established by Sakurai et al. (2007). The direction of ball spin was expressed by three coordinate values and spin rate was defined by the absolute value of the angular velocity vector. A helical axis of rotation was calculated and expressed as two separate angular deviations from the x-axis, across two different planes (Fig 2a). The oblique angle (θ) was measured as the spin axis' deviation from the x axis in the x-z plane. The elevation angle (ø) was measured from the horizontal (or x-axis), in the x-y plane (Fig 2b). The angle between the velocity vector of the ball and its axis of rotation was calculated as the alpha angle (α). Digital video footage from the posterior camera was analysed using Siliconcoach software (Dunedin, New Zealand). The maximum lateral deflection of the ball was measured as the horizontal distance from the most extreme point of deviation in the ball's trajectory to the point where it crossed the goal line.

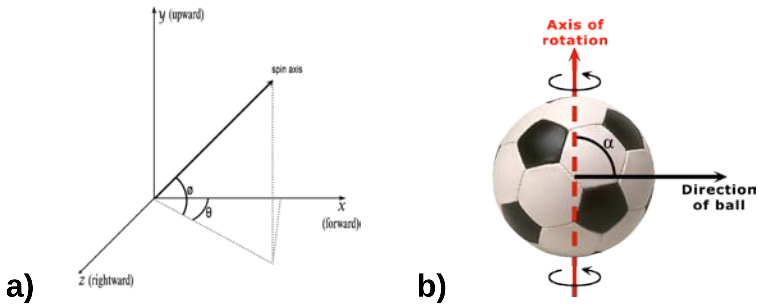

Figure 2. Illustration of a) the measurement of the ball rotation axis in three dimensions and b) the angle (α) between the velocity vector of the ball and its axis of rotation.

As this was a pilot test, no inferential statistics were run; however, a bivariate correlation was used to establish the association between, ball spin and ball velocity, and ball spin and lateral deviation.

RESULTS: As would be expected, the angular velocity of the ball was significantly higher in the C (36.4 rad.s^{-1}) compared with S kick (22.6 rad.s^{-1}), whereas velocity was similar.

Table 1. Mean, standard deviation and range of selected ball flight variables (n=4).* significant (p<0.05)

	Mean	S.D.	Range
Ball Rotation (rad.s^{-1})			
Straight Kick	22.6	4.5	3.1 – 4.7
Curved Kick	36.4	5.1	4.8 – 6.6
Ball Velocity (m.s^{-1})			
Straight Kick	22.1	2.9	18.3 – 25.4

Curved Kick	19.2	2.1	16.9 – 21.2
Alpha angle (α) (deg)			
Straight Kick	93.0	14.1	78.3 – 110.9
Curved Kick	85.3	12.8	66.2 – 93.0
Oblique angle (θ) (deg)			
Straight Kick	85.4	16.0	62.5 – 99.8
Curved Kick	95.2	24.3	62.0 – 113.4
Elevation angle (ø) (deg)			
Straight Kick	30.4	18.7	14.8 – 57.5
Curved Kick	62.6	3.9	56.9 – 65.3
Lateral Deflection of ball (m)			
Curved Kick	1.49	0.08	1.37-1.55

The elevation angle (ø) of the spin axis was significantly greater in the C (62.6°) compared with S kick (30.4°). The spin axis remained relatively perpendicular to the velocity vector of the ball, irrespective of the kick being performed (α - S: 93.0°; C: 85.3°). A strong negative correlation (r=-0.81) between ball velocity and spin rate existed, whereas a strong positive relationship (r=0.89) was found between ball angular velocity and lateral deflection. Collectively, the rotational characteristics above resulted in the ball deviating 149.3 cm from the lateral extreme of its parabolic trajectory, to where it crossed the goal line.

DISCUSSION: Whilst the respective linear foot velocities were not significantly different between kicks, ball velocity was approximately 3 m.s^{-1} lower in the C kick. As the ball is kicked off-centre to produce rotation, it is logical to assume that there is a decreased transfer of energy through the centre of mass of the ball, explaining the lower ball velocity seen in the C kick. Kicking literature states that linear foot velocity is the best predictor of ball velocity (Lees & Nolan, 1998) however; it would appear that this theory may be less applicable to C kicks. The product of linear foot velocity and offset distance of impact would be an appropriate predictor of ball velocity in C kicks. The high spin rate in the C kick was to be expected. The explanation as to why such a noticeable angular velocity (22.6 rad.s^{-1}) existed in the S kicks most likely lies in the contact point on the ball. This point of contact seemed to lie inferior to the horizontal midline of the ball and as a result, backspin was imparted to the ball. The angular velocity seen in the S kick was not unexpected, as it is reasonably impossible to apply zero eccentric force to a ball when kicking. Contacting the ball even slightly off-centre with as much force, as is seen in a kick for distance, will decisively cause a degree of ball rotation. The spin axis remained relatively perpendicular to the velocity vector of the ball, irrespective of the kick being performed. In the transverse plane, the spin axis was essentially aligned with the goal line in both kicks. The slight differences seen in this oblique angle can be explained by considering the contact point on the ball. In the S kick, the ball was aligned with the target area, so the kicker merely had to contact the inferior aspect of the ball at its midline, to send it directly at the target. In the C kick however, the kicker had to initially direct the ball slightly away from the target area, in order to avoid the obstacle and let the curvature in its trajectory bring it back on target. In order to direct the ball in such a manner, the player kicks *across* the ball, again contacting the inferior aspect of the ball but this time to the right of its midline. The reactive motion of the ball to this horizontally *and* vertically aligned off-centre impact is enough to shift the oblique angle of the spin axis and account for the 10 degree difference. The most noteworthy difference in the rotational axis of the ball was its elevation from the horizontal (Fig 3). In the S kick, the participants seemed to be attempting to position their kicking foot as parallel to the ground as possible, so as to expose a large, flat surface to contact the ball slightly below its horizontal midline (which, in theory, would create a horizontal spin axis). However, such a foot position is mechanically impossible to achieve when kicking, so the foot was actually held oblique to the ground at impact, thus explaining the slightly elevated spin axis (~30°). This contrasts with the C kick, where the elevation angle of the spin axis was more than twice that seen in the S kick (62.6°). This is explained by the kicker contacting the ball off-centre, thus creating a horizontal torque around a more vertical axis. This axis orientation seems logical as the sideways force acting on a spinning sphere acts perpendicular to the velocity vector and the spin axis (Mehta, 1985). Therefore an upright spin axis would enhance lateral deflection. The increased elevation angle in the C kick allows the ball to deflect more horizontally and since the players were trying to curve the ball from right to left in the horizontal plane, a vertical rotational axis would be the most efficient way to achieve their goal. Obviously creating a perfectly upright axis of rotation is unattainable when kicking, but the increased elevation of the spin axis seen in the C kick is proof of the kickers' intent.

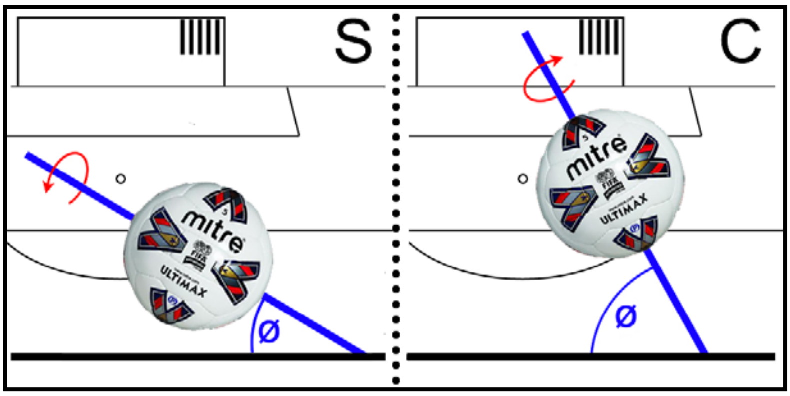

Figure 3. The elevation angle (ø) in the straight (30.4°) and curved (62.6°) kicks.

The strong negative correlation (r=-0.81) between ball velocity and spin rate supports the notion of a trade off existing between the two which has previously been alluded to in the literature (Asai et al., 2002). However, maximising ball spin may not be in the best interests of a player, as the ensuing velocity trade off would provide the goalkeeper with more time to react to the kick. The strong relationship between ball spin and lateral deflection (r=0.89) is supported in the literature.

CONCLUSION: Ultimately, it would appear that it is not in the kicker's best interests to apply excessive eccentric force to the ball when taking a free kick. Instead, the velocity-ball spin trade off suggests that players should aim to create no more ball rotation than is necessary for their given kick. This would reduce the detriment on ball velocity and ensure the ball reaches the goal as quickly as possible. The ball should be directed as close to the edge of the defensive wall as possible. Even in C kicks, kicking with power must be a primary goal. From the perspective of the goalkeeper these findings have direct implications regarding the positioning of their defensive wall. Many goalkeepers use the imaginary line drawn between the ball and the nearest goalpost to align their wall. They will direct the outermost player on the defensive wall to stand along this line, effectively blocking any straight access to the goal for the kicker. This study suggests that it is not beyond high level kickers to curve the ball around this player and into the goal, at high velocities. Therefore, as a rule, we suggest that for free kicks within 25 yards (23 m) of goal; the second-to-outermost player in the wall should stand on this imaginary line. This will increase the deviation required for a successful kick by ~50-60 cm, and in turn the ball rotation required to achieve this deviation. This will negatively affect ball velocity and allow the goalkeeper more reaction time to save the ball.

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TASK DECOMPOSITION AND THE HIGH PERFORMANCE JUNIOR TENNIS SERVE

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To develop consistency in the toss placement and racket trajectory, coaches often decompose the serve and practise it in separate parts. This study compared the kinematics of the ball toss as part of the discrete serve skill and when the skill was decomposed. A 22 camera VICON MX motion analysis system, operating at 250 Hz, captured racket and ball kinematics of 5 elite junior players hitting flat first serves (FS) directed to the 'T' of the deuce service box and a ball toss (BT) drill where players were instructed to perform the decomposed skill as in the FS. Paired t-tests were used to assess within-group differences. Vertical displacement of ball zenith increased significantly (~20cm) during BT. Consistency in select racket and ball kinematics characterised the FS, while this appeared to decrease in BT.

KEYWORDS: skill development, whole-part practice

INTRODUCTION: The first serve, the most important stroke in tennis is also the stroke that has attracted the most investigative interest from biomechanists. The kinematics of lower and upper limb joint motion as well as the movement of the racket have been examined (Fleisig et al., 2003; Chow et al., 2003), while other researchers have preferred to explore the relationships between select joint kinetics and joint injury or the skill's performance (Elliott et al., 2003; Reid et al., 2007). Given that successful serve performance is ultimately governed by impact between racket and ball, it's surprising that so little research has attempted to comprehensively evaluate this link.
Traditionally, tennis coaches have emphasised the need for mechanical consistency in stroke production and therefore the performance of the serve. This approach, however, contrasts with more contemporary principles of skill acquisition, where variable movement patterns are considered functional facets of performance (Davids et al., 2001). In an effort to simplify the learning of the serve or reduce its dimensionality, coaches often decompose the service action or, in coach parlance, break it down into its component parts. Similar findings have highlighted the link between perception and action in cricket batting, where the temporal and kinematic features of a forward defensive drive change when a batsman faces a ball machine instead of a bowler. Indeed, the ability to nurture information-movement coupling has been proposed to rely heavily on the specificity of training (Savelsbergh and van der Kamp, 2000) and therefore questions the efficacy of skill decomposition in skill development.
The aims of this study are therefore (a) to investigate the hand, ball and racket kinematics of the flat serve (FS) of elite junior players and (b) to examine the effect of a ball toss drill, where the toss is rehearsed independent of the swing (BT), on those kinematics.

METHOD*:* Five nationally-ranked male right-handed junior players aged 13.40 ± 0.54 yrs and 164.86 ± 8.46 cm tall participated in the study with their own racket. Following a standardised warm-up, participants were instructed to perform two tasks, the first being to hit 10 successful, maximal effort FSs at a 1 x 1 m target area bordering the T of the service box on the deuce court. Ten trials were selected to attain a statistical power of 90% with only 5 participants (Bates et al., 1992). The second task focused on the performance of the BT component of the FS skill. Players were instructed to toss the ball as they would in a FS but without racket-ball contact; thus rehearsing or simulating the ball toss of a FS. All participants indicated that they were familiar with the isolated performance of the BT, having routinely engaged in this type of drill during practice. Each participant completed the tasks

(FS and BT) in a randomised manner, with a two minute rest period permitted after each block of 10 serves.

A 22-camera, 250 Hz VICON MX motion analysis system (Oxford Metrics Inc., UK) was used to track the 3D marker trajectories.The marker set consisted of; 3 retro-reflective markers on the left hand, one marker on the head of the first metatarsal (LMT1) of the left foot, 5 markers on the racquet and 3 markers on the ball. All displacements were made relative to each player's foot position by re-positioning the origin of the global coordinate system to the position of the LMT1 marker (Figure 1). LMT1 position was determined during each participant's address and prior the players initiating the backswing of their serve. Within the global co-ordinate system, positive X was to the right; positive Y was forward and positive Z upward.

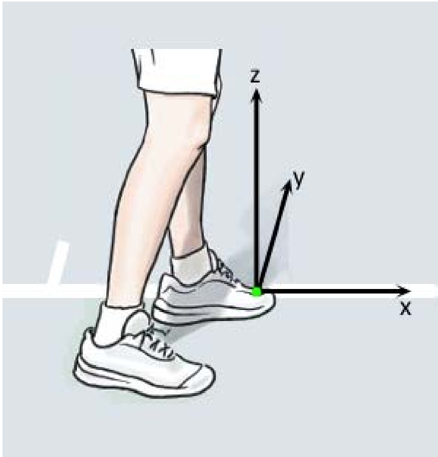

Figure 1. Position of global coordinate system.

Impact was determined as one frame (0.004 s) prior to racket-ball contact. In the BT condition, this was accomplished by determining the mean vertical displacement of the ball at impact during the FS. In each BT trial, the frame in which the ball best approximated the mean vertical displacement recorded in each subject's FS was considered to represent impact. Gaps in the marker trajectories were filled using the cubic spline interpolation function within VICON Nexus. To account for impact accelerations, a second order polynomial extrapolation was performed in line with previous recommendations (Knudson and Bahamonde, 2001). All data subsequent to one frame pre-impact were deleted before the extrapolation was performed using customised Matlab software (The Mathworks, Natick, Massachusetts, USA). A Woltring filter with an optimal mean squared error (defined as 25 mm) was applied to the raw data. Raw anatomical, racket and ball data were modelled using the University of Western Australia's customized full body, racket and ball models, respectively. A Z-X-Y order of rotation was used to express the rotational axis of the ball and calculate rotation rate. Thirteen paired t-tests were used to investigate if kinematic differences existed between FS and BT conditions. Due to the multiple comparisons being conducted, statistical significance was adjusted a priori to $p<0.01$.

RESULTS: All reported data are for the mean (± SD) for the five subjects over the 10 trials. In the FS condition, ball position at height of the toss (BZ) trended further forward and to the left, while its vertical displacement was significantly higher ($t=-5.794$, $p=0.004$) in the BT condition (BT: 311.2 ± 24.4 cm; FS: 288.2 ± 19.2 cm). The position of the ball at impact did not change significantly in any plane. Ball rotation also increased significantly in the BT condition (FS: 837 ± 343 vs BT: 927 ± 333 deg.s^{-1}; $t=-4.893$, $p=0.008$).

Table 1. Comparison of hand, racket and ball kinematics in the FS vs BT conditions.

	Flat Serve (FS)		Ball Toss (BT)		
	Mean	S.D.	Mean	S.D.	P
x ball pos @ zenith (cm)	-17.8	9.1	-8.0	13.1	.111
y ball pos @ zenith (cm)	39.6	11.1	35.3	13.1	.444
z ball pos @ zenith (cm)	288.2	19.2	311.2	24.4	.004*
x ball pos @ imp (cm)	-34.0	9.5	-22.8	12.6	.189
y ball pos @ imp (cm)	51.8	8.5	40.8	16.0	.246
z ball pos @ imp (cm)	250.9	9.3	250.8	9.31	.458
Ball rot during toss (deg.s^{-1})	837	343	927	333	.008*
Toss time (s)	0.802	0.079	0.886	0.122	.246
Release to zenith (s)	0.526	0.042	0.568	0.043	.015

Zenith as % of toss	66.0	1.71	61.8	1.43	0.001*

Toss time increased ~10% from 0.802s in the FS to 0.886s in the BT. BZ when expressed as a percentage of toss time occurred significantly earlier when the skill was decomposed (FS: 66.0 ± 1.71 v BT: 61.8 ± 1.43%, t=8.918, p=0.001). Only the vertical displacement (and resulting flight time characteristics) of the tossed ball changed significantly between the two conditions. Table 2 presents the variability of the 3D ball position at BZ in the FS compared with BT conditions. This was achieved by calculating the mean and standard deviations of the standard deviations describing the x, y and z displacement at BZ (Davids et al., 2001). The position of the ball at BZ trended toward being most variable in the lateral direction during the FS. When the ball toss was practised in isolation, the consistency of the lateral (FS: 5.49 ± 1.19 cm vs BT: 7.76 ± 2.40 cm; t=-2.969, p=0.041) and vertical (FS: 5.06 ± 1.82 cm vs BT: 9.00 ± 3.43 cm; t=-2.924, p=0.043) positions of ball zenith appeared to deteriorate.

Table 2. Comparison of the mean and standard deviations of 3D ball position at BZ in the FS and BT.

	Flat Serve (FS)		Ball Toss (BT)		
	Mean	SD	Mean	SD	p
x ball pos @ BZ (cm)	5.5	1.19	7.8	2.40	0.041
y ball pos @ BZ (cm)	10.1	7.70	8.2	2.47	0.648
z ball pos @ BZ (cm)	5.1	1.82	9.0	3.43	0.043

DISCUSSION: While the position of the ball at BZ was further forward and to the left, only its vertical position changed significantly when the toss was rehearsed independently. The ball was tossed ~24 cm higher when the skill was decomposed; logically accounting for the increased toss time observed in this condition (FS: 0.802 s; BT: 0.886 s). These results would appear contrary to coach suggestions that isolating a component of the skill makes it easier for the player to replicate that component. The changes in both the spatial and therefore temporal aspects of the toss suggest that the players tossed the ball in a manner notably different to that seen during their FS. These results are consistent with previous work that has illustrated the deleterious effect of decoupling the toss from the swing in the volleyball serve (Davids et al., 2001). While it could be argued that the effect of this type of task decomposition would be less pronounced among elite adult performers, the investigated sample have practised their serves extensively and were familiar with the task performed. As compared to professional players directing first serves to a similar court location, these players made racket-ball impact further to the left but not as far forward (Chow et al., 2003).

The ball's average peak rotation during the toss increased significantly during BT (FS: 837 deg.s^{-1}; BT: 927 deg.s^{-1}). This is instructive as it points to players applying more force to the ball, not just in the vertical direction to bring about the increase in BZ, but also eccentrically. In practice, the authors have observed coaches encourage players to rehearse the BT to reduce spin rates – the very characteristic that it would appear to be amplifying in this group of players. One of the primary motives behind decomposing a highly organised skill is to condense the informational load to augment learning (Naylor et al., 1963). For this reason, it was with interest that there was a trend for ball placement to become more variable in the forward and vertical directions during the BT drill. Inadvertent introduction of larger amounts of variability in the vertical position of BZ may impair rather than assist a movement pattern, which has been refined around a lower and relatively (or at least *more*) stable BZ.

The small sample size is considered a limitation of this study and future research is encouraged to further explore the effects of task decomposition or whole-part practice on tennis stroke biomechanics.

CONCLUSION: Consistency in select ball toss kinematics characterise the performance of the FS from a young age. This consistency decreases when the serve is decomposed, as is routinely done by coaches, while key characteristics of the serve, like BZ, change significantly when the ball toss is practised in isolation. These differences point to the role of information-movement coupling in the serve and question the efficacy of practices that involve the decoupling of ball and swing.

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THE SURFACE EMG ACTIVITY OF THE UPPER LIMB MUSCLES IN TABLE TENNIS FOREHAND DRIVES

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The purpose of this study was to analyze the 3D kinematics variables and the upper limb muscle surface EMG activity of Taiwan elite table tennis players when they were performing forehand drives after receiving topspin and backspin services. Ten Vicon MX-13^{+} cameras (Vicon, Oxford, UK, 250Hz) were used to record the 3D kinematics data and measured the EMG signals of seven upper limb muscles of the players. The results showed that the tactics of the table tennis players performed the forehand drive to receive backspin were both to increase the racket tilt angle in advance and to raise the path angle during the upswing phase. The players exerted greater muscular activity during receiving the backspin forehand drive than receiving topspin forehand drive in the wrist extensor, the biceps and the triceps.

KEYWORDS: biomechanics, IEMG, topspin, backspin.

INTRODUCTION: Table tennis is one of the most popular racket sports in the world. The forehand drive is one of the most classical and effective technique of the table tennis skills. When the table tennis players perform forehand drive, they will meet several situations, including the topspin and the backspin shots from the opponent and so on. The comparison of forehand drives between receiving topspin and backspin is a great topic that the table tennis players are interested. Previous studies of table tennis focused on 3D kinematics and EMG methods to describe the motions of table tennis forehand strokes. This includes the studies such as, Kasai, & Mori (1992) described the movement appearance of the forehand table tennis drives. Neal (1991) he found the elite Chinese players performed the faster initial velocity of ball than the Australian young players. Only a few researchers analyze the movement of table tennis strokes with the methods of EMG. Yoshida, Sugiyama, & Murakoshi (2004) observed the muscular EMG activity patterns of the table tennis forehand shots. They found that the EMG patterns and the movement duration time were similar in different forehand drives while returned the different spin and found that the duration time from the ball rebound on the table to the contact point of the forehand drives were about 0.2 seconds. The purposes of this study were to compare the kinematics variables and the EMG signal patterns of the forehand drives when the players were receiving the topspin and the backspin table tennis services.

METHOD: Five male table tennis elite players in Taiwan (with an average age of 22.6±3.36 years, height of 175.2±6.14 cm and weight of 66.2±13.21 kg) served as the participants. Figure 1 shows the schematic drawing of the experimental setup. The players were standing at one end of the table to return the services. The opponent server served the topspin and the backspin services into the circle (25cm) on left end corner of the participant player's. The players moved to the left side to play a straight forehand drive into the 50×50cm square at right end of the opponent. The 3D kinematics data were recorded by using ten VICON Motion Capture systems MX13+ (250 Hz) of forehand drives, and the Vicon Nexus 1.4 software was used to calculate the kinematical parameters. One Biovision EMG system (1000Hz, Biovision, Wehrheim, Germany) was to collect the EMG signals of seven upper limb muscle groups, which were the wrist flexor, wrist extensor, biceps brachii, triceps brachii, pectoralis major, deltoid and trapezius. The EMG data were analyzed by using the Acknowledge software (1000Hz). Raw EMG signals were band-pass filtered (20-500Hz) and the full wave rectified by passing it through a linear envelope with a window of 10 ms. The EMG signal of the muscle was standardized by the peak amplitude of each muscle during the experiment. The integrated EMG (IEMG) signals from the preparation phase, the contact point to the follow through phase were analyzed. The sequence of the EMG signal activities, the EMG amplitude at the contact point, the peak EMG amplitude and the IEMG of the upper limb muscle groups during different movement phases were the selected variables. The kinematics, the standardized EMG and IEMG of the selected muscles were tested between the forehand drives after returning topspin and backspin services by the Wilcoxon matched-paired signed rank nonparametric statistical test. All the variables were tested by SPSS 18.0 statistical software at a 0.05 significant level.

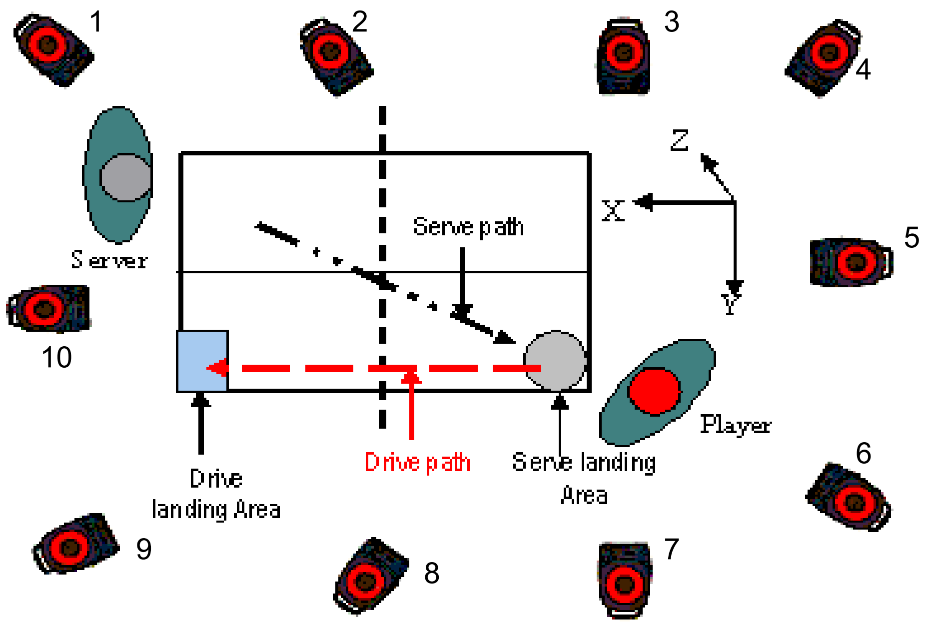

Figure 1 · The Schematic of the Experimental Setup

RESULTS: Figure 2 and figure 3 show the rectified EMG signal patterns of forehand drives by one of the subjects. The lines in the figure 2 and the figure 3, the line 1 means the start of the downswing, 2 means the end of the downswing and the start of the upswing, 3 means the contact point, 4 means the end of the follow through. The phases of the forehand drive were divided into the downswing phase (1 to 2), the upswing phase (2 to 3) and the follow through phase (3 to 4). Table 1 shows the kinematical data of the different forehand drives. Table 2 shows the EMG variables of every muscle group.

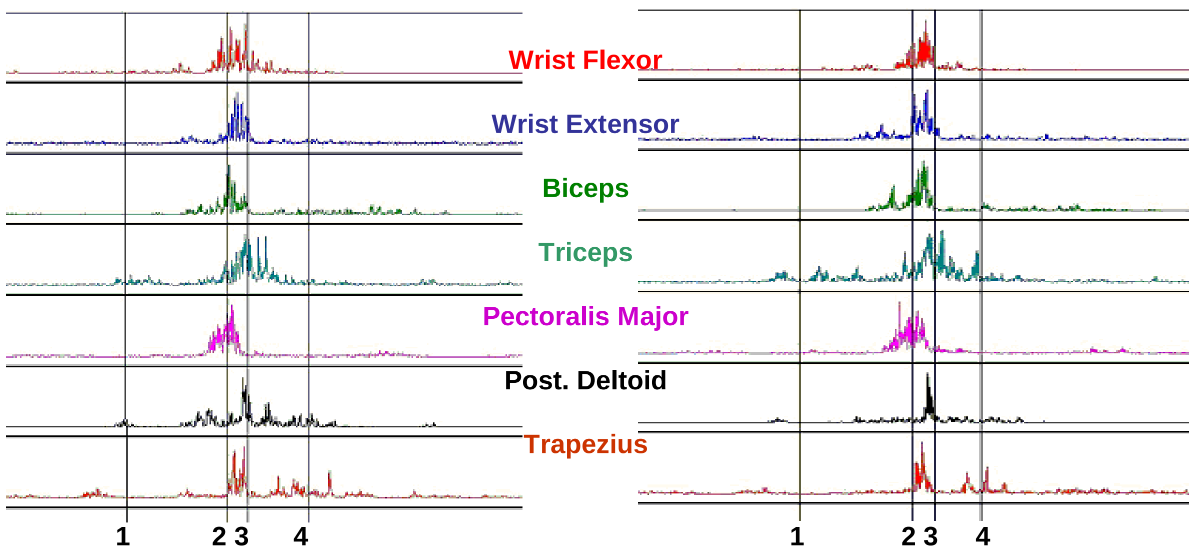

Figure 2. The EMG of receiving topspin drive **Figure 3. The EMG of receiving backspin drive**

Table 1. The Kinematics Variables of Different Forehand Drives

Variables	Receive Topspin Drive	Receive Backspin Drive	*p*
Racket Head Velocity at Contact *(m/s)*	17.31±1.12	18.94±1.70	*
Racket Saggital Tilt Angle *(deg)*	54.84±2.47	70.72±3.34	*
Racket Up Swing Angle *(deg)*	33.12±11.07	42.98±8.27	*
Contact Height *(m)*	0.95±0.03	0.91±0.04	*
Total Movement Time (TMT) *(s)*	0.85±0.10	0.92±0.03	
Down Swing Duration Time *(s)*	0.50±0.07	0.60±0.08	*
Down Swing Duration Time / TMT*(%)*	58.0±5.96	65.2±6.30	*
Up Swing Duration Time *(s)*	0.09±0.02	0.08±0.02	
Up Swing Duration Time / TMT *(%)*	11.0±2.12	9.2±1.92	*
Follow Through Duration Time *(s)*	0.22±0.06	0.18±0.03	
Follow Through Duration Time / TMT *(%)*	25.0±5.66	20.2±4.09	*

p<0 .05

Table 2. The EMG Variables of Forehand Drives

Muscles	Shots	EMG at Contact (%)	*p*	Peak EMG (%)	*p*	Peak EMG Timing (s)	*p*	Down Swing IEMG (%s)	*p*	Up Swing IEMG (%s)	*p*	Follow Through IEMG (%s)	*p*	Total Motion IEMG (%s)	*p*
Wrist Flexor	Topspin	13.87		85.27		-0.043		3.23		2.22		2.08		7.61	
	Backspin	19.62		84.01		-0.027		3.35		2.33		1.69		7.46	
Wrist Extensor	Topspin	32.33		74.23	*	-0.070		2.42	*	2.61		1.47		6.56	
	Backspin	20.74		97.76		-0.067		3.18		2.80		1.25		7.30	
Biceps	Topspin	18.92		81.31	*	-0.043		2.62		2.32		1.06	*	5.23	
	Backspin	16.97		100.00		-0.039		2.12		2.64		0.59		5.40	
Triceps	Topspin	23.36		68.97	*	0.007		1.89	*	2.11		2.89		6.99	*
	Backspin	22.65		96.07		-0.008		2.80		2.65		2.80		8.31	
Pectoralis Major	Topspin	3.80		74.89		-0.089		2.49		1.97		0.65		5.15	
	Backspin	4.32		73.09		-0.103		3.02		1.85		0.77		5.67	
Deltoid	Topspin	16.10		83.17		-0.076		2.66		1.88		1.89		6.51	
	Backspin	18.72		86.12		-0.036		2.43		2.26		1.40		6.14	
Trapezius	Topspin	12.13		87.15		-0.021		2.19		2.35		1.28		5.88	
	Backspin	9.59		82.54		-0.072		2.48		2.00		0.91		5.46	

p< 0.05

DISCUSSION: Table 1 showed that the racket head velocity of receiving backspin forehand drive (18.94 m/s) was significantly faster than the receiving toping drive (17.31 m/s). The saggital tilt angle of the receiving backspin forehand drive (70.72 deg) was significantly greater than that of the receiving topspin drive (54.84 deg) at contact. And the swing path angle of receiving backspin forehand drive (42.98 deg) was significantly greater than the receiving topspin drive (33.12 deg). The contact height of receiving backspin forehand drive (0.91 m) was significantly lower than that of the receiving topspin drive (0.95 m). The total movement time (TMT) between the different drives was 0.85 second vs. 0.92 second, there was insignificant difference between them. But the percentage of the movement phases were all different, we found that the receiving backspin forehand drive would spend a longer time to the downswing phase to prepare, the receiving backspin serve forehand drive spent a shorter period of time in the upswing phase. Figure 2 and figure 3 showed the EMG amplitude rose from the end of downswing movements. The sequences of two drives were not consistent from the central muscle group to the end of the segment muscle groups. Table 2 showed that there were insignificant differences in the EMG amplitude at the contact point and the IEMG during the upswing phase. The peak EMG amplitudes of upper limb muscles appeared just before the contact point, except the triceps in the receiving topspin serve drive. During the downswing phase of the action, the IEMG signal was different between two drives in the wrist extensor and the triceps muscles. The players increased the racket tilt angle in advance just before the upswing movement. The triceps exerted the greater IEMG signal in receiving backspin serve forehand drive than in receiving topspin forehand drive. That might be the fact that the triceps was to apply a brake in counteracting the upswing movement.

CONCLUSION: In this study, we combined the 3D kinematics and EMG methods to compare the sequence muscular activity, EMG amplitude and IEMG signal of upper limb muscles between two different table tennis forehand drives while received the topspin and the backspin serves. We found that the racket head velocity of receiving backspin forehand drive was greater than the receiving topspin forehand drive. The players performed the receiving backspin forehand drive in a longer downswing duration time and a shorter duration time of the upswing. The tactics of the table tennis players to perform the forehand drive in receiving backspin would increase the racket tilt angle in advance and increase the upswing path angle. The players exerted greater muscular activity in the wrist extensor, the biceps and the triceps during receiving the backspin forehand drive than receiving topspin forehand drive.

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AN INVESTIGATION OF SOCCER BALL VELOCITY ON INSTEP KICK WITH AND WITHOUT ARM SWAYING

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KEYWORDS: Kinematics, powerful kick, football

INTRODUCTION: Many biomechanical researches of soccer try to understand the kinematic and kinetic effects of kicking; the most widely concerned is maximal force instep kick (powerful instep kick). Kellis and Katis (2008) reported that the ball velocity in instep kick on related researches were surpass in 72 Km/h (20m/s), and the highest was 115.6 Km/h (32.1 m/s). The higher ball speed, the lower time a keeper react. The kicking skills are complicated, and multiple factors must be measured. Not only lower limbs involve on kicking task, but also upper body affects the kicking performance. Kicking with arm swaying is a common motion in goal kick, directly free kick, penalty kick, and instep kick. The arm (non-kicking side) begins to sway when a player run-up to kick a ball at the last step and it may provide more strength to kick a ball. Arm swaying motion also applies in other sports. It advances the jumping performances in standing long jump due to maintaining balance and increasing the velocity of the body's center of gravity (Ashby & Heegaard, 2002). The muscle pre-lengthening and stretch by arm swaying increased the strength of kicking; skilled soccer players had significantly larger knee flexion-extension movement (Shan and Westerhoff, 2005). The range of motion (ROM) of knee and the foot segment velocity are the important parameter for understanding the skill of players. The distance between right ankle and left ankle (while the support-foot landing) was measure to understand the hip and body stretch, and this distance affect the time of acting force for kicking the ball. To improve the ball velocity on kicking, the purpose of this study was to compare the ball velocity and kinematic data on different motions of arm. The motions were arm swaying and arm fixed (without arm swaying) while kicking the ball. The results were a useful reference for soccer players and instructors.

METHOD: An elite football player serviced in national team (age: 25 years-old, height: 175 cm, weight: 80 Kg, right foot dominant), without lower extremity injuries within six month volunteered to participate in this study. A motion capture system with 10 cameras (sampling at 250Hz, VICON MX13^{+}, Oxford Metrics Ltd, England) was used to collect kicking data. A radar gun (sampling at 300 Hz, STALKER Ltd, USA) was used to measure the peak ball velocity (FIFA approved Adidas soccer ball, model: final8, size: 5, Pressure: 0.6 bar). The force plate (sampling at 1000Hz, model: 9187, KISTLER, Switzerland) was used to define the landing time of support-foot during instep kick. The kicking tasks were arm swaying and arm fixed on instep kick. A self-selected approach angle (30 degree) and step (2 steps) by participant were used for the test trial, and following trial were succeed the same angle and step, which would allow a maximal instep kick. The participant was asked to sway the left arm (non-kick side) over 90 degree (shoulder extension) and arm fixed (kicking with arms akimbo) to kick the ball 10 times separately with instep kick (as fast as possible) into the target net (width: 1m and height: 0.8 m, the distance from ball to target net was 1.5 m), then the data were collected. All subjects were tested using a full-body plug-in-gait marker set. The Visual3D V4.0 software (C-motion Inc, USA) was used to calculate the foot velocity, the range of motion (ROM) of knee, ROM of shoulder, and the distance between right ankle and left ankle (while the support-foot landing). The descriptive analysis was used to analyze kinematic data.

RESULTS: The data of ball and foot velocity and the ROM of knee and shoulder are shown in Table 1. The distance between right ankle and left ankle was 98.1±2.1cm (arm swaying) and 93.8±2.7cm (arm fixed). The instep kicks with arm swaying had the higher ball velocity, foot velocity, knee movement, and right and left ankles distance then arm fixed. The range of ball velocity on arm swaying was 94~100 Km/h, and arm fixed was 90~95 Km/h.

Table 1. The ball and foot velocity and ROM of knee and shoulder

	Ball velocity (Km/h)	Foot velocity (m/s)	ROM of knee (degree)	ROM of shoulder (degree)
Arm swaying	96.87±2.03	13.60±1.07	92.80±6.43	177.90±1.77
Arm fixed	93.00±1.41	13.07±0.61	86.40±3.65	27.25±1.74

DISCUSSION: In this study, there had two reasons for causing the higher ball speed in instep kick with arm swaying. The first was the arm swaying provides more range of motion on hip and knee joint, thus the ROM of knee and kicking distance were greater than that without swaying in instep kick. The second was instep kick with arm swaying produced more strength or strength translation to the ball due to muscle pre-lengthening or the greater balance controlled on body. Thus, the kicking leg had the better position to kick the ball. Finally, the results could reliably be used to evaluate a player's kicking ability (Shan and Westerhoff, 2005).

CONCLUSION: Base on this study, the arm swaying advanced the effective instep kick. The instep kicks with arm swaying caused the higher ball velocity and body stretch. This investigation may provide a useful reference to coaches or teachers for approach optimal instep kick. Future studies could describe the functions of arm swaying during instep kick on motor control perspective.

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SCREENING TEST FOR THE POTENTIAL RISK OF ACL RUPTURE OF FEMALE AND MALE SOCCER PLAYERS

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ACL rupture is a devastating injury. An analysis of risk factors and a consecutive prevention program might help to reduce the risk of injury. However there exists no screening test to identify the individual ACL-injury risk. The purpose of this study was to develop a screening test by which the potential risk of ACL rupture for female and male soccer players can be estimated. Testing procedure focussed on dynamical knee valgus in frontal plane during landing of a drop jump in normal and fatigued state. The results were obtained by 2D video analysis. Results show a wide range of dynamical knee valgus in both sexes with a 5.4 cm greater dynamical valgus of women emphasizing the distinctly higher risk potential for ACL injury of women. Screening tests seem to be suitable to achieve an estimation of the individual risk for ACL injuries without large expenditure.

KEYWORDS: ACL, prevention, soccer, fatigue, screening test.

INTRODUCTION:

Rupture of the anterior cruciate ligament (ACL) is a devastating knee injury for the athlete with a long phase of rehabilitation and inability to perform sports. Additionally it may cause degenerative changes of joint. Most frequent situations causing ACL ruptures (review in Alentorn-Geli et al. 2009a) are landing positions after a jump, or stopping, plant and cut manoeuvres. High risk sports are team handball, soccer and basketball. 70 to 84% of ACL injuries occur in non-contact situations. Studies have shown a 2.4 to 9.5 higher incidence of ACL rupture in women than men. Commonly risk factors have been divided into environmental, anatomical, hormonal, neuromuscular and biomechanical - simply into intrinsic (within the body) and extrinsic (outside the body) - factors. Quatman & Hewett (2009) emphasize the evidence of the “valgus collapse” mechanism in frontal and sagittal plane as strong predictor of ACL-injury risk for females. Mandelbaum et al. (2005) discussed the frontal plane dynamic knee valgus during landing in drop jump as potential risk. Prevention programs for soccer and team handball focussing on modifiable intrinsic risk factors (biomechanical and neuromuscular) have been developed mainly showing positive results (review in Alentorn-Geli et al. 2005b). Nevertheless the effectiveness of prevention programs is not yet regarded as assured (Bahr & Krosshaug 2005). Karlsson (2010) postulates the implementation of preventive measures in normal sports training on an every day basis. But there exist no screening test to identify the individual risk of ACL injury and no method to monitor training induced alterations. Such a testing could have an essential advantage to analyse individual factors of the athletes and it could contribute to the development of individual prevention programs. Therefore it was the aim of the present study to develop a simple screening test for to estimate the potential risk of ACL rupture of female and male soccer players.

METHODS:

Two junior soccer teams (U17) of both sexes, 15 female players (15.5 ± 1.1 yrs , 65.9 ± 8.7 kg, 170.1 ± 6.0 cm) and 15 male players (15.7 ± 0.7 yrs, 71.7 ± 7.2 kg, 180.5 ± 8.4 cm) without acute injuries or former ACL ruptures participated in the study. After a 5 minute warm-up on a cycle ergometer an isometric strength test was carried out (Cybex norm) consisting of 3 MVCs of knee flexor (45° knee angle) and extensor (90° knee angle) muscles in randomized order. Next the young players performed 3 drop jumps from a height of 40 cm and also of 60 cm (Huston et al. 2001) in rested and - after an exhausting stepping exercise - in fatigued state. Finally a second isometric strength test was carried out. For the right and the left leg two 3-dimensional force plates (Kistler) were used to measure ground reaction forces with a sampling frequency of 1000 Hz. Two video cameras (JVC, 50 Hz) in frontal and sagittal plane were used to determine dynamic knee valgus and knee angle between the first contact to ground and the deepest bending position. Video and force plate data were synchronised and stored with the software Simi-Motion. 2D-analysis was performed by Simi Twinner Pro. In frontal plane horizontal distances between left and right knee (knee distance, KD) as well as between left and right foot at ankle height (foot distance, FD) at ground contact as well as in the deepest bending position were measured by 2D analysis, also knee angle in sagittal plane. A position with a lower KD than FD is defined as a valgus position (VP). A decreasing knee distance between ground contact and deepest bending position is defined as dynamic knee valgus (DKV). The results were processed by the means of Simi-Motion, Simi-Onforce and Excel. Furthermore SPSS V.17 was used for statistical analysis.

RESULTS: Results of isometric MVCs (tab. 1) show (except female right leg) a significant reduction of quadriceps force (♀ -4.4%, ♂ -7.0%) after exhausting exercise, but no significant changes in hamstring force. Male players show significant higher forces (absolute: quadr. 51%, hamstr. 43%; in relation to body weight: quadr. 38%, hamstr. 32%) under all conditions.

Table 1. Isometric MVC [Nm] of quadriceps (quadr.) and hamstrings (hamstr.) in rested und fatigued state, t-test (p).

		quadriceps		Hamstrings		t-test (p)	
		Normal	**fatigue**	**Normal**	**Fatigue**	**quadr.**	**hamstr.**
female	**left**	151.1 ±29.5	140.9 ±20.1	110.0 ±14.4	113.7 ±11.3	**0.028**	0.070
	right	152.1 ±27.7	149.1 ±25.5	109.5 ±25.7	115.8 ±18.8	0.252	0.058
male	**left**	230.6 ±51.0	219.9 ±49.3	155.5 ±32.1	163.2 ±27.2	**0.034**	0.061
	right	232.1 ±58.7	210.5 ±37.3	162.8 ±31.6	162.2 ±27.8	**0.015**	0.439
t-test (p) sex	**left**	**0.001**	**0.000**	**0.000**	**0.000**		
	right	**0.001**	**0.000**	**0.000**	**0.000**		

Results of drop jumps (tab. 2, fig. 1) show the same FD (28.6 ±3.8 cm) for male and female players in the moment of first ground contact at landing. KD of women at ground contact is at least significantly lower (-3.0 ±2.9 cm) than FD. In contrast KD of men does not show a significant difference to FD (-1.5 ±2.5 cm). Up to the deepest bending position, women show a significant increased VP (-7.9 ±4.8 cm) synonymous to a DKV. In contrast the slightly increased VP of men (-2.4 ±4.8 cm) is not significant. Additionally there is a significant greater jump height of men (♂ 20.7 ±6.1 cm, ♀ 17.2 ±3.7 cm). The force results of drop jumps related to body weight show no significant gender differences.

Table 2: FD, KD and VP at landing (ld) resp. at deepest bending position (dp) and DKV during drop jumps (dj) from a height of 40 cm and 60 cm in rested und fatigued (f) state, t-test (p)

		horizontal distance [cm]			valgus [cm]		t-test (p)	
							FD	**KD ld**
	dj	**FD**	**KD ld**	**KD dp**	**DKV**	**VP dp**	**KD ld**	**KD dp**
Female	**40**	28.9 ±4.9	25.3 ±3.0	21.0 ±4.1	-4.3 ±3.7	-8.0 ±5.2	**0.020**	**0.003**
	40f	28.4 ±3.6	25.5 ±3.0	20.0 ±4.1	-5.5 ±4.4	-8.4 ±5.2	**0.034**	**0.001**
	60	29.0 ±3.4	25.6 ±2.3	21.4 ±3.0	-4.2 ±3.5	-7.6 ±4.7	**0.004**	**0.000**
	60f	28.0 ±3.6	25.0 ±3.3	20.4 ±5.7	-4.6 ±4.6	-7.7 ±5.2	**0.012**	**0.000**
Male	**40**	28.8 ±3.1	27.2 ±2.7	26.3 ±4.8	-0.8 ±4.1	-2.5 ±5.2	0.134	0.558
	40f	27.5 ±4.5	27.1 ±3.3	25.3 ±6.2	-1.8 ±4.5	-2.2 ±5.0	0.792	0.323
	60	29.2 ±3.5	27.2 ±2.4	27.0 ±3.1	-0.2 ±3.8	-2.2 ±4.7	0.129	0.907
	60f	28.9 ±4.1	27.0 ±3.3	26.1 ±5.3	-0.9 ±3.8	-2.8 ±4.2	0.178	0.587
t-test (p) sex	**40**	0.921	**0.030**	**0.003**	**0.009**	**0.006**		
	40f	0.304	0.201	**0.007**	**0.004**	**0.000**		
	60	0.898	0.237	**0.006**	**0.017**	**0.010**		
	60f	0.751	0.146	**0.005**	**0.020**	**0.016**		

At ground contact KD of women is only during drop jumps from 40 cm height in rested state significant lower than KD of men, whereas in deepest bending position women show significant higher VP and DKV than men (tab. 2, fig. 1). In total 95% of women and 65% of men show an increased valgus during landing and therefore a dynamical knee valgus. Defining a DKV of at least 5 cm as criterion of an increased risk of ACL injury, 75% of women and 32% of men show an increased risk, equivalent to a 2.5 higher risk of women than men.
Comparison of sorted single values of women and men show similar results (fig. 2) independent from jump height or fatigue. Women and men show a similar range and distribution of DKV with an about 5.4 cm higher valgus of women.

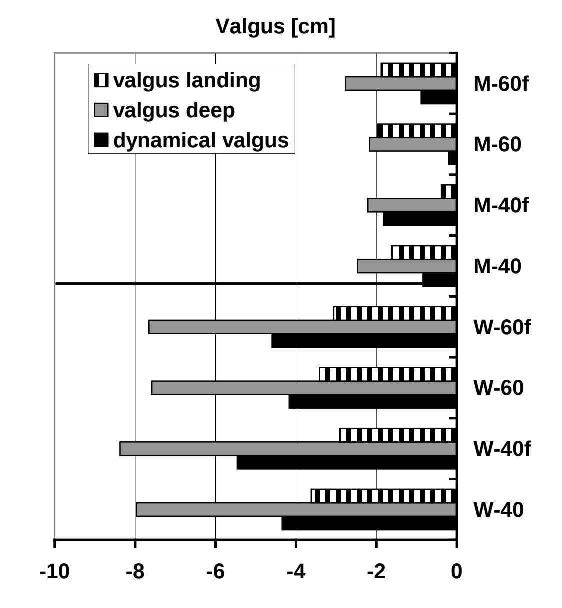

Figure 1: Valgus at landing and at deepest bending position, DKV during landing of women (W) and men (M) during drop jumps from a height of 40 cm and 60 cm in rested and fatigued (f) state (s. tab. 2).

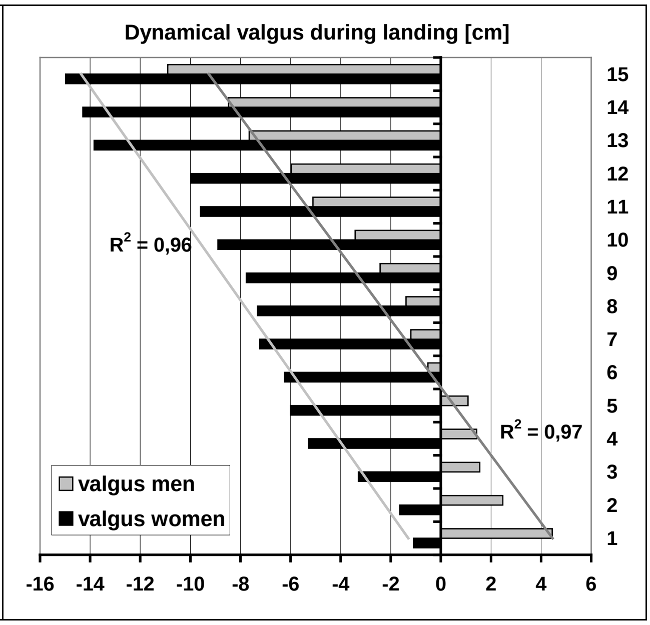

Figure 2: Sorted data of DKV [cm] of women and men during landing, individual mean values over all conditions, sorted by valgus.

Results of maximum knee angle while landing show no significant differences between women and men neither in rested (40 cm: ♀ 77.8 ±9.8°, ♂ 81.7 ±10.9°) nor in fatigued state (40 cm: ♀ 77.9°±8.7°, ♂ 83.1 ±11.3°) nor dependent on drop jump height. Higher drop jump height leads to a significant higher knee angle in both sexes (60 cm: ♀ 82.6 ±13.6°, fatigued 83.0 ±12.8°; ♂ 89.0 ±13.2°, 88.1±11.4°).

DISCUSSION: Results of MVCs show a significant but only slight reduction of quadriceps force after exhausting exercise. One reason may be the time shift of at least 15 min. between the jump test and the first MVC caused by the positioning and fixing procedure on the isokinetic testing station (cybex norm). The unchanged MVCs of hamstrings indicate that stepping was not suitable for fatigue of hamstrings. Therefore a possible fixing of tibia by hamstrings against shifting and protection of ACL was not affected by the exhausting exercise.
The study shows no dependences between DKV during landing and drop jump height or fatigue. Therefore it seems to be suitable to perform a screening test to estimate the potential risk for ACL injury with only one drop jump height of about 40 cm without fatigue.
The leg positions during landing verify the anatomical caused tendency of increased valgus deformity of females (Alentorn-Geli et al. 2009a). The higher degree of DKV of females during landing after jumps, emphasized as strong predictor of ACL-injury risk (Mandelbaum et al. 2005, Quatman & Hewett 2009), is obvious. In Addition results underline the assumption

that an increased initial valgus at ground contact also leads to a higher degree of DKV during landing - caused by working forces and torques. Against most literature (cf. Alentorn-Geli et al. 2009a) both sexes show no difference in knee angles while landing. If this could be confirmed in further investigations, a screening test might be performed without respect to knee angle.
The arbitrarily defined criterion - in the absence of an existing threshold level - of at least 5 cm dynamical valgus as indicator for a potentially higher risk of ACL injury is still unclear and has to be verified in future works. However with this assumption results show a 2.5 higher risk of females, which is in the same order as the higher risk factor of ACL injury described in literature (Alentorn-Geli et al. 2009a). Furthermore results show not only in 75% of women but also in 32% of men an increased risk of ACL injury, which was not to be seen from the group mean values. Particularly, the high range and the similar distribution of DKV seem to be an interesting finding. The higher potential risk of women is expressed by the on average 5.4 cm increased DKV. Nevertheless, it becomes quite clear that the risk of ACL injury is not limited to female athletes. In consequence an ACL prevention program (e.g. FIFA „11+“, Grimm & Kirkendall 2007) for athletes of both sexes with a higher risk potential is recommended.

CONCLUSION: The presented video-based screening test is a first step in order to achieve important hints at the potential ACL injury risk without large expenditure. It supports the hypothesis of Karlsson (2010) to implement preventive measures in normal sports training on every day basis. Identifying and monitoring ACL injury risk seems to be important regarding the fatal results of ACL injury for the athletes. Further work is to be done to verify and to refine the screening test. Here one should focus on threshold value for the DKV, sex specifity or individual foot position. All in all, it is important to investigate whether and, in particular, how prevention programs reduce the individual risk level.

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ISBS 2010

Oral Session 11

Balance

A CASE STUDY ON BALANCE RECOVERY IN SLACKLINING

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The purpose of this study was to identify and describe the basic balance recovery movements performed during slackline balancing. Slacklining is an activity where the athlete balances on a thin piece of webbing that is mounted between two fixed points in a not too tight way. We designed an experimental setting where a controlled perturbation is applied to the slackline and study the movements of athletes to regain a balanced position. Four athletes took part in the study and for each we recorded five trials using a Vicon motion capture system. With the help of a 15 segment biomechanical model we studied mechanical quantities like the center of mass trajectory, the energy contributions, and also analyzed joint actuation patterns.

KEYWORDS: balance, slackline, motion control

INTRODUCTION: Slacklining is an activity where an athlete balances on a 2.5-5 cm wide piece of nylon webbing stretched between two anchor points. In contrast to tightrope walking, a slackline is only moderately tightened and allowed to stretch and bounce. Setups include “normal” lines of roughly ten meters that are close to the ground, long lines of up to 250 meters, high lines up to 1000 meters over ground and jump lines that allow tricks like rotations and flips.
The purpose of this study is to explore the underlying processes that allow the athletes to stabilize their bodies in this highly unstable setting.

METHOD: The research questions arising are manifold. How does the subject deal with perturbations on such a slackline? Which joints are actuated to regain a stable position? How far can the center of mass be deflected from the (unstable) equilibrium on the line before the subject falls? How much energy can be dissipated?
In order to address those questions a slackline (with 6.9 meters length, 0.46 meters above ground, with a pre-tension of 4000 Newton, leading to a maximum slack of 0.3 meters) was set up in a local gym. The line was laterally deflected at 0.53 meters from one fixed point using a ratchet system. The subject places itself on the slackline such that it is unable to anticipate the release of the deflection mechanism. After reaching a balanced position the deflecting rope is cut and a lateral perturbation is introduced into the system. The balance recovery movements are then measured and analyzed. Four slackliners with medium to very high skill levels took part in this experiment, each of which performed five trials.
We use a Vicon V612 motion capture system with eight near-infrared cameras and 100Hz sampling frequency. The subject is equipped with 55 spherical (14mm diameter), retroreflective markers. Another 12 markers are mounted on the slackline and the deflection mechanism and an additional 3D accelerometer (sampling at 100Hz) is also mounted on the slackline at the application point of the perturbation.
In order to determine body segment inertia parameters 95 anthropometric measures are taken according to Yeadon (1990) and joint coordinates are determined using a Global Optimization approach (Lu & O'Connor, 1999). Measurement artefacts are reduced by applying a modified zero-lag Butterworth filter with 10Hz cut-off frequency.

RESULTS: In general, two slightly different behaviours are shown by the test persons. While two of them balanced with two feet on the slackline before introducing the perturbation the other two balanced on one foot only. Results in this section are concentrated on the latter method, but do generally also agree with the former one.

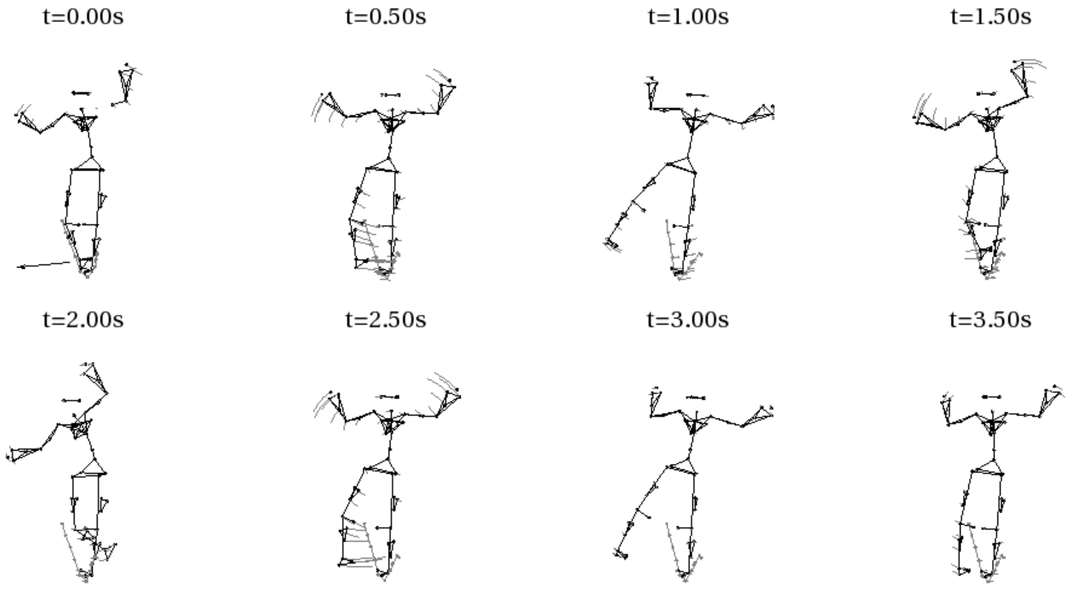

Figure 1. The reconstructed positions of markers on the subject and the slackline at times ranging from t=0s to t=3.5s viewed from the back for trial A. The marker trajectories during the preceding 0.2s are also included. At t=0s the line is released and moves in the direction of the arrow to the original position.

In Figure 1 marker positions after the release of the deflection mechanism (t=0) are shown in 0.5s intervals, where the athlete is viewed from behind. In this trial (denoted trial A) the subject experiences a perturbation in the line such that the right foot is pushed to the left. The center of mass (CM) is no longer close to the equilibrium point. In order to regain a balanced position the subject has to rearrange the CM, the support point on the line (in this case the right foot) and the line's fixed points in a plane spanned by the vector between the anchor points and the vector of the gravitational force. A clockwise rotation of the whole body is seen within the first second. Several oscillations of the body are seen thereafter. About four seconds after release the subject has regained a configuration where only small correction are needed to stay on the slackline. Acceleration peaks of 35 m/s^2 were recorded over the first 0.2 s and the slackline travelled 5 mm after release. The same information is shown for an unsuccessful trial (denoted trial B) in Figure 2.

In Figure 3 the center of mass (CM) component in x direction (perpendicular to the line and parallel to the ground) as well as the x-component of the center of the supporting foot is displayed against the time after release for the successful trial A (left figure) and the unsuccessful trial B (right figure). In the successful trial the leg undergoes rapid oscillations, which decrease in magnitude after around two seconds. In the unsuccessful trial the CM is displaced too far from the equilibrium position in the beginning of the counter movement, which leads to a fall shortly thereafter. Note that in trial A the trajectory of the supporting foot oscillates roughly around the CM path, which leads to forces in both directions, while in trial B no equilibrium restoring forces occur.

The total energy of the subject (i.e. kinetic and potential contributions) is plotted for trial A and B in Figure 4. In the successful trial A a peak in energy appears shortly after releasing the displacement mechanism and is dissipated in consecutive steps. After three such oscillations we arrive at an energy level comparable to the level before the perturbation. On the contrary for trial B a sharp rise in energy appears shortly after applying the perturbation and cannot be dissipated anymore.

Uncertainties in the results arise mainly from fluctuations in the reconstructed marker trajectories, soft tissue movement artefacts and uncertainties in anthropometric quantities and the applied model. In Figures 3 and 4 a band of uncertainty is included in the plot that is determined from measurement fluctuations, while we expect soft tissue movement contributions to be suppressed due to the use of the Global Optimization approach. Modelling uncertainties still need to be determined.

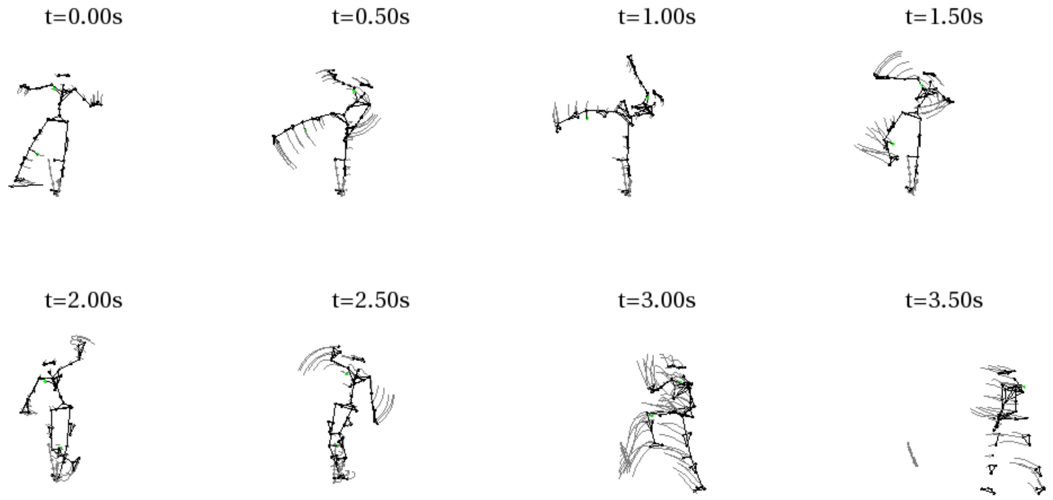

Figure 2. The reconstructed marker positions on the subjects and the line for the unsuccessful trial B.

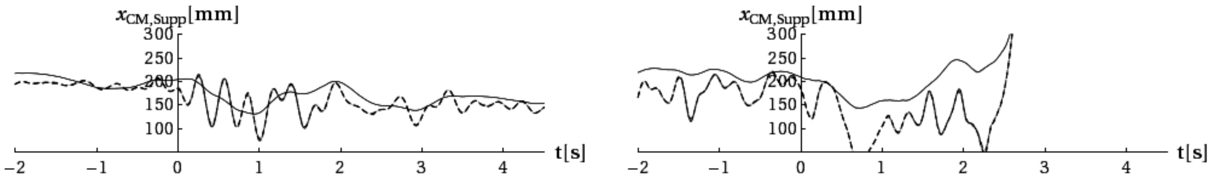

Figure 3. The x-components of the center of mass trajectory (solid line) and the supporting foot on the line (dashed line) for trials A (left) and B (right). An error band is included for both quantities, but it is hardly visible due to the small relative uncertainties of length measurements.

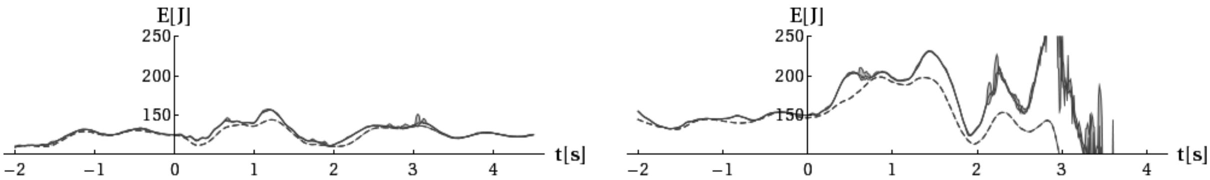

Figure 4. Potential energy (dashed line) and total energy (solid line, kinetic plus potential energy) of the subject plotted against the time after release for trial A (left) and trial B (right). In trial B the decrease in energy after 3.5 s appears because the subject had to step off the line. Note that the measurement uncertainties described above are displayed in form of an error band, which is mainly visible in the latter phase of trial B.

DISCUSSION: The presented results were collected in a pilot study with only a small group of experienced slackliners. Nevertheless, the basic mechanism of balance recovery movements can be identified. An equilibrium configuration of the system athlete/slackline is the combination where the two fixed points of the line, the support point of the body on the line and the athlete's CM are in the plane spanned by gravity and the fixed points. If the CM lies outside this plane there is a force driving the CM away from the equilibrium point. A change in body configuration and the elastic properties of the slackline allow the athlete to apply a force and angular moment that can eventually restore the equilibrium position. Figures 1 and 2 suggest that the main contributions to this change in body configuration is due to movement in the arms. Depending on the technique applied also the non-supporting leg is used to balance.

Comparing the x-components of the CM trajectory and the supporting leg also provides an important insight to the mechanism of balance recovery. In order to experience a restoring force from the line it is necessary to assume a configuration where both, the CM and the

support point are displaced in the same direction normal to the plane described above. Due to the small damping in the system multiple oscillations of the line appear before a stable region is approached. Furthermore, none of the subjects managed to correct situations, where a lateral distance of more than 100mm between CM and the contact point on the line was measured.
After the release of the deflection mechanism the total energy of the subject increases. Although even with additional kinetic energy the athlete can stay on the line (which is also called “surfing” the line due to the swinging motion) it is easier to react to consecutive perturbations with minimal kinetic energy. The dissipation of energy is observed in an oscillatory way, which might be due to the elastic properties of the line.
On a different note, any basic knowledge on movement regulation patterns of healthy, well trained subjects under these highly unstable conditions may help to better interpret human balance regulations in general as well as balance disorders.
Slacklining could also improve balancing capabilities in other sports and therapy. A possible way to study effects would be to measure and analyze balancing capabilities in quiet standing (e.g. by means of nonlinear analysis, Cagran et.al. (2010)) before and after basic slackline training.

CONCLUSION: We present results of a pilot study to address the mechanism underlying balance recovery on a slackline. A lateral perturbation is introduced into the system and the reaction of the test person was studied. Basic mechanical quantities, e.g. the center of mass trajectory and energy contributions were computed. As expected from experience all subjects predominantly used their arms to react to displacements of the center of mass. Measurements also show that energy due to the counter movements is dissipated in a stepwise fashion. Also none of the subjects were able to regain balance when the center of mass was further than 100 mm away from the contact point on the line in lateral direction.

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EFFECT OF COMBINED LOCAL TOPICAL ANESTHESIA AND PHYSICAL ACTIVITY ON KNEE PROPRIOCEPTION SENSES, AND STATIC BALANCE IN HEALTHY YOUNG INDIVIDUALS

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Sixty males participated in this investigation and were randomly assigned into four groups. Group one received only local topical anesthesia on the dominant knee. Second group performed 10 min. sub-maximal running on treadmill with speed of 8 km/h. Third group received local topical anesthesia, the same as group one, combined with 10 min. sub-maximal running. Fourth group served as a control group. The knee proprioception senses for active and passive repositioning, and quadriceps, hamstring maximal torque were measured using Biodex Isokinetic Dynamometer. A single leg stance balance test was used to measure static balance. Findings revealed two interesting results. First; 10 min. sub-maximal running improved knee passive repositioning; second; after 10 min. sub-maximal running static balance declined.

KEYWORDS: proprioception, static balance, topical anesthesia.

INTRODUCTION: The complexity of joint motion requires joint stability especially in the movement which sudden and fast change in joint motion is necessary. Joint stability combined with smooth coordinated movement requires synchrony of agonist and antagonist muscles contraction. Proprioception information is collective neural input from joint capsules, ligaments, muscles, tendon, skin, and mechanoreceptors which provide neuromuscular coordination to protect the joint from injury, by using the appropriate balance of synergistic and antagonistic forces. Proprioception information also have important role to maintain good balance and posture. Balance requires proper information from sensory input, effective processing by the CNS, and appropriate responses of motor control (Kauffman et. al, 1997). The CNS relies on information from three sensory systems for maintaining balance. Proprioception, visual, and vestibular. input provides information about the orientation of the body and body parts relative to each other and the base of support. Information is received from joint and skin receptors, deep pressure and muscle proprioception (Hobeika, 1999; Horak, 1994). Proprioception cues are dominant inputs for maintaining balance when the support surface is firm and fixed (Hobeika, 1999; Dietz 1980).
Muscle spindles have been regarded as the most important among proprioceptors, however in recent year's skin mechanoreceptors role became increasingly apparent (Edin, 2001). Neurophysiologic evidence that afferent information from skin receptors is important for proprioception has been gathered mainly in experiments relating to the human hand and finger joint (Collins et al., 2005). Receptors in the hairy skin of humans can provide high-fidelity information about knee joint movement (Edin, 2001). Clinical data suggest that information from skin mechanoreceptor is valuable for joint stability. Physical therapists have claimed that taping a joint improves its stability. Taping large joints such as the knee hardly makes any mechanical support. However the ability of the organism to control the muscle acting on the joint may be due to altered somatosensory inflow from the skin (Edin, 2001).
Sport injuries are unwanted part of the game. Topical anesthesias have been often used to numb the pain of injured limb in order to help the athlete to continue participation in the game. Topical anesthesia is a liodcane analgesic medication which is easily absorbed by the skin and provides local numbness to the area of application. Warm up before physical activity improve proprioception (Subasi et al., 2008) while fatigue decreases propriocetion capability (Surenkok et al., 2006). It is unclear if using topical anesthesia followed with physical activity which simulates real game situation would have impact on receptors in the skin, proprioception, and static balance. The purpose of this study was to assess the effect of combined local topical anesthesia and physical activity on knee proprioception senses, static balance, and quadriceps and

hamstring maximum torque.

METHOD: Sixty male physical education students (age 22.1±3.6 years, height 177.4±5.4cm, weight 70.7± 10.6 kg) from participated in this investigation. Subjects were free from history of orthopedic problem in their dominant knee and ankle. They randomly assigned into four groups. Group one received only local topical anesthesia 10 cm below and 10 cm above and around dominant knee. Second group performed 10 min. sub maximal running on treadmill with speed of 8 km/h. Third group received local topical anesthesia the same as group one combined with 10 min. sub maximal running. Fourth group served as a control group. All subjects participated in pre and post tests. The knee proprioception senses for active and passive repositioning, and quadriceps, hamstring maximal torque were measured using Isokinetic Dynamometer Biodex System 3.Single leg stance balance test was used to measure static balance. Pared t -test and ANOVA was used to analysis the data.

RESULTS: The results are shown in Table 1. No significant differences were found in the knee proprioception senses, static balance and maximum quadriceps and hamstring torques post application of local anesthesia. After 10 min. sub maximal running the knee proprioception sense for passive repositioning improved significantly, while balance declined, and no significant differences were detected for quariceps and hamstring maximum torque. Combined application of local topical anesthesia and sub-maximal running improved knee proprioception for passive repositioning, but decreased static balance; and showed no significant effect on quadriceps and hamstring maximum torque. No significant differences were found between pre and post tests for knee proprioception, static balance, and quadriceps and hamstring maximum torque in control group. (P value of 0.05 was chosen).

Table 1. Pared t-test comparison between 3 groups

	Group 1 (topical anesthesia)		Group 2 (10 min running)		Group 3 (combined anesthesia & running)	
	t	p	t	p	t	p
Active joint	**-0.08**	**0.39**	**-0.69**	**0.49**	**-1.6**	**0.12**
Passive joint	0.07	0.94	2.69	0.018*	2.6	0.02*
Static Balance	-0.47	0.64	2.99	0.01*	2.4	0.02*
Quadriceps Maximal	-0.49	0.69	-0.048	0.96	-0.21	0.83
Hamstring Maximal	-1.8	0.08	-1.2	0.25	1.3	0.82

*(P<0.05)

DISCUSSION: Findings of this study revealed two interesting results. First; 10 min. sub-maximal running improved passive knee repositioning, while active repositioning did not change significantly. Muscle spindles are the first source of proprioception senses (Riemann et. al 2002); and there is a possibility that their roles are more important in passive than active movement (Subasi et. al 2008). Thus 10 min. sub maximal running could have acted as a medium to warm up the muscles and caused higher sensitivity of muscle spindles. Second; after 10 min. sub-maximal running static balance declined. In order to control balance one is more dependent on information from the lower leg and ankle musculature. In this case lower leg and ankle muscles fatigue might have been cause of static balance decline (Miura et. al 2004).

Application of local topical anesthesia did not show significant effect on knee proprioception and single leg stance balance. Skin mechanoreceptors are important in the human hand (Edin& Johansson, 1995) and face (Gracco & Abbs, 1985). Mechanoreceptors in joint ligaments and capsules may be crucial in extreme, but not necessarily noxious, joint position (Burke et al.1988; Edin, 1990). Temporally skin numbness by local topical anesthesia may be more effective to block pain receptors than the skin mechanoreceptors. There is also possibility that in gross movement of large joint such as knee, muscle spindle receptors are dominant and skin mechanoreceptor numbness has no impact on knee proprioception.

CONCLUSION: Based on the finding of this study local topical anesthesia has no effect on knee proprioception senses, static balance, and quadriceps and hamstring maximum torque. 10 min. sub maximal running improved knee proprioception sense but decreased static balance which could have been due to lower leg and ankle muscles fatigue.

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THE EFFECT OF KOREAN FOLK DANCE EXERCISE TO THE KINEMATIC PARAMETERS FOR DOWN STAIRCASE WALKING OF ELDERLY PEOPLE

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KEYWORDS: elderly, stair walking, descending, folk dance, lower extremity, kinematics

INTRODUCTION: According to the Statistics department of Korea in 1991, population aged over 65 years in 1970 was 991,000 people (3.1% of whole population) and it has increased to 3,370,000 people (10.7% of whole population) in 2000. By 2010 Korea will become a full-scale aged society with an estimated population of about 7.53 million, 13.7% of those being the elderly. This phenomenon was occurred primarily due to the fertility reduction by the family planning project which was held since 1962 and extension of average life expectancy. Compared to the foreign countries which manage the aging population over long periods of time, the growth of the elderly population in Korea was faster than any other country in the world. Thus, the preparation for this is very urgent matter. Aging changes their movements and activities of people. Physical and functional degradation due to the aging process affect muscle atrophy, muscle loss, and loss of muscle function (Schlicht, Camaione & Owen, 2001). Muscle weakness affects the balance keeping which is an important factor to cause a falling. Falling is the most common injury of elderly people and even a light fall seriously hurts the elderly and can often lead to death (Leibson, Toteson, Gabriel, Ransom & Melton, 2002). Elderly's falling occured more often to elderly women than men and 60% of this accident happens during daily activities. Startzell et al. (2000) also indicated that stair decending falls occured three times more often than for stair ascending. Walking downstairs is not only an essential motion but also is the most fearful action for elderly's daily life process. However, most previous research on the characteristics of stair walking were targeted to normal young people. Thus, in this study, Korean folk dance program as part of rhythmic movement exercise was applied to the elderly women for increasing dynamic balance, stalility and range of motion, and the determined effect of it will be analyzed selected kinematic parameters.

METHOD: Twenty elderly women (70$\pm$1.3 yrs, 154.9$\pm$9.2 cm, 63$\pm$14 Kg) were recruited as the participants. Kinematic data were collected by seven real-time infrared cameras (125Hz, Vicon, England) while subjects walked stair descent as a pre-test. Thirty seven reflective makers were attached to the body for 3D-analysis and COM and joint angels of lower extremity at each event were calculated. The Korean folk dance exercise program was practiced by the elderly for 12 weeks. All participants performed this dance exercise for 50 minutes a day, three times a week. Same experiment on stair descent walking was performed as post-test. A stride of gait was divided into five events such as E1: initial contact of 1st foot, E2: mid-stance at maximal vertical GRF, E3: toe-off, E4: mid-swing with maximal knee angle and E5: initial contact of 2nd foot. For comparison of kinematics at each event, a paired t-test was performed to observe if significant differences existed between pre and post test (p<0.05).

RESULTS: The COM of anterior-posterior (A/P) direction shows significant differences between pre-and post tests at E4 and 5 (table 1). Knee angles also show significant differences between the tests for flexion/extension at E1, and for internal/external rotation at E2, E3, and E5 (table 2).

Table 1. COM (unit:m)

		E 1	E 2	E 3	E 4	E 5
M/L	Pre	0.20(0.07)	0.20(0.04)	0.20(0.06)	0.19(0.05)	0.20(0.04)
	Post	0.20(0.04)	0.21(0.04)	0.20(0.01)	0.20(0.04)	0.19(0.05)
	t	0.130	0.406	0.102	0.602	0.819
A/P	Pre	1.56(0.07)	1.49(0.06)	0.99(0.09)	0.71(0.09)	0.40(0.22)
	Post	1.52(0.07)	1.40(0.08)	1.00(0.11)	0.82(0.09)	0.66(0.13)
	t	0.906	1.661	0.138	3.154*	3.646*
Ver	Pre	1.24(0.05)	1.20(0.05)	1.04(0.05)	0.98(0.09)	0.89(0.05)
	Post	1.24(0.07)	1.20(0.09)	1.06(0.10)	1.00(0.07)	0.93(0.11)
	t	0.194	0.050	0.249	1.357	1.247

Note. * significant difference at $p<0.05$

Table 2. Knee Angle (unit:degree)

		E 1	E 2	E 3	E 4	E 5
M/L	Pre	2.37(10.46)	14.84(5.34)	41.48(34.34)	48.15(28.60)	8.39(5.07)
	Post	13.21(8.60)	19.59(11.51)	56.10(31.30)	65.72(27.39)	27.42(33.56)
	t	5.360*	1.192	0.728	1.202	1.573
A/P	Pre	-0.62(6.71)	0.08(10.72)	-4.20(25.61)	5.29(33.38)	0.86(7.35)
	Post	-3.82(16.82)	1.22(12.51)	5.88(25.80)	1.83(30.37)	-6.70(23.45)
	t	0.775	0.342	1.745	0.523	1.223
Ver	Pre	-4.25(8.62)	-11.98(14.75)	-5.34(8.92)	-12.62(16.71)	-12.39(8.67)
	Post	-2.39(4.94)	1.78(5.10)	13.28(11.72)	9.44(17.57)	-4.76(5.17)
	t	0.393	2.921*	2.813*	1.731	2.427*

Note. *significant difference at $p<0.05$, M/L: flexion/extension, A/P: varus/valgus, Ver: internal/external rotation

DISCUSSION: Cesari (2005) mentioned that old adults at descent stairs preferred lower staircase height due to the less flexibility of lower extremity compared to the young adults. In this study, after the 12 week exercise treatment, elderly subjects showed significant increase of COM in the anterior direction during swing phase. During the stride, kinematic data also indicated that knee is more flexed and rotates internally after the treatment. These results reflected that the exercise treatment may affect to increase ROM of knee joint and improve the gait pattern of stair descent.

CONCLUSION: This study proved that Korean folk dance can be a tool of training elderly people for improving dynamic balance and mobile abilities. It is expected to be able to contribute to the welfare of the elderly in an aging society by utilizing this program and also to develop fall-related injury prevention program.

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BIOMECHANICAL ANALYSIS OF TAI CHI CHUAN FIXED-STEP PUSH-HAND

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KEYWORDS: Tai Chi Chuan, Push-hand

INTRODUCTION: Tai Chi Chuan is a Chinese traditional martial art. It is not only helpful to health (Frye, Scheinthal, Kemarskaya, and Pruchno, 2007) but also a fighting skill. After routine training for a long time, Tai Chi Chuan learners will do an advance sparring set training, that is, Push-hand. They are trained to use their tactile sense well and apply the 'Eight Methods of Tai Chi' to attack and defend in reality (Wu, 1994) in Push-hand process. Chan, Luk, & Hong (2003) revealed the kinematic characters while a master performing the push movement in Tai Chi. Most of the principles of Push-Hand were only recorded in Tai Chi Chuan ancient books and records. Consequently, the purpose of this study was to identify biomechanical characters in Push-hand process and to expound it scientifically.

METHODS: Six male subjects (28.4±1.64 yrs, 175.2±5.42 cm, 64±7.95 kg) practicing Tai Chi Chuan at least five years were divided to three couples in this study. Subjects were asked to do fixed-steps(not moving steps) Push-hand with single hand operation – Peng style, and to put each of their feet on a force plate. There were totally 67 reflective makers placed on one subject. And these makers were placed on joints to establish spatial coordinate systems on segments. Human body was divided to fifteen segments by markers in this study, and segments were thought as rigid bodies.The ground reaction force (GRF) and reflective markers` spatial coordinates data for both subjects during performing fixed-steps Push-hand were collected by KISTLER force plates and VICON motion analysis system respectively. All data were analyzed by Visual 3D and MATLAB softwares and filtered by 6Hz low-pass filter.The motion of distal segment relatives to proximal segment, that is, joint angle were described in Euler`s angle, and the rotation sequence was flexion–extension (x), abduction–adduction (y), external rotation-internal rotation (z). Furthermore, origional three dimensional GRF data in the laboratory coordinate system were transformed to the local body coordinate systrem for describing personal motion consistently. GRF values were normalized to body weight (BW), and COM displacement were normalized to body height (BH).

RESULTS: All subjects performed the same characters in Push-hand processes. The trajectory of COM was a smooth and repeating oval trajectory, and the COM of one subject moved forward while another moved backward (figure 1). But the COM undulated small in vertical direction. Responding to the trajectory of COM, the greatest GRF on back (front) foot appeared at the maximal backward (forward) displacement of COM. In additional, maximal rotation of the waist was after the maximal backward displacement of COM and the flexion/extension motion of the waist were small. The flexion/extension motion of elbow was also revealed. Table 1 show the motion range of waist and elbow for once attack and defence of the Push-hand process. And table 2 show the GRF peak values for once attack and defence in the whole Push-hand process.

Table 1. Motion ranges of human kinematic parameters

parameter	angle
Rotation of waist	**8.28 ± 3.14**
Flexion-extension of waist	**4.65 ± 2.34**
Flexion-extension of elbow	**34.83 ± 14.3**

Table 1. GRF peak values

GRF	min (BW)	max (BW)
Back foot - Vertical	**0.277 ± 0.071**	**0.833 ± 0.056**
Front foot - Vertical	**0.161 ± 0.056**	**0.696 ± 0.086**
Back foot - Forward	**0.046 ± 0.028**	**0.104 ± 0.045**
Front foot - Backward	**0.02 ± 0.025**	**0.13 ± 0.109**

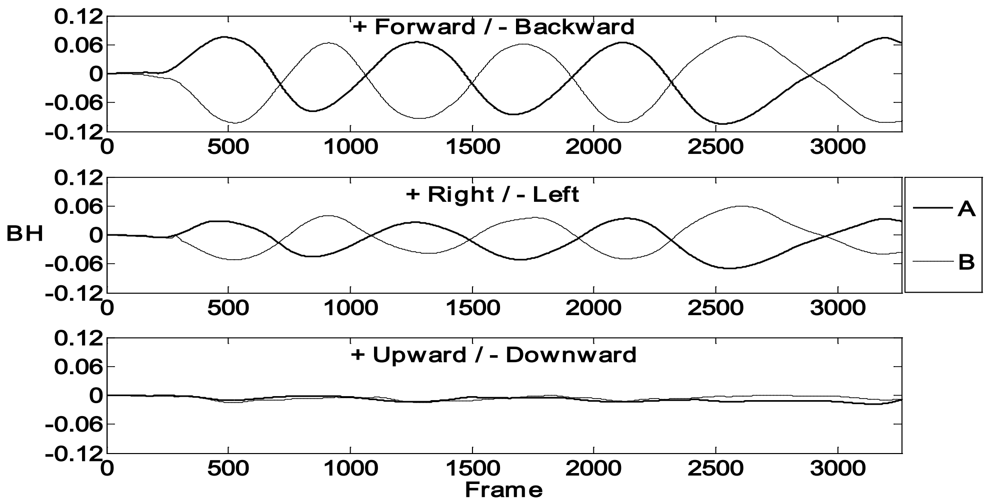

Figure 1. COM trajectories of one couple subjects. A is one subject, and B is another.

DISCUSSION: The COM trajectories showed that subjects attacked with COM moving forward. A subject needed to move backward to ward off another's attack. The movement of COM did not rely on moving steps and flexing the waist. In additional, the rotation of the waist was more important. The small flexion-extension range revealed a principle of Tia Chi Chuan, that is, keep an upright posture of the trunk. And the small displacement of COM in vertical indicated the same purpose, that is, keep balance. The GRF varied with COM motion. It was always greater than half body weight on one foot at the maximal COM displacement. These results were responded to Tai Chi Chuan theories. The flexion-extension motion of elbow was not small and it conflicted to the principles of Tai Chi Chuan. It was considered about the level of subjects.

CONCLUSION: Characters in Push-hand process were identified. The GRF and the trajectory of COM varied with each other. Additionally, the motion of COM did not only rely on rotation of waist but also combined motion of joints of lower limbs. Consequently, the joint motion of lower limbers will be investigated in future.

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AN ARTIFICIAL NEURAL NETWORK METHOD FOR PREDICTING LOWER LIMB JOINT MOMENTS FROM KINEMATIC PARAMETERS DURING COUNTER-MOVEMENT JUMP

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The purpose of this study was to develop an artificial neural network (ANN) model for predicting the joint moments of lower limbs using solely the kinematic parameters during counter-movement jump (CMJ). Nine female volleyball players performed CMJ. The joint moments were calculated from experimental data by inverse dynamics (called "measured joint moments" in this study). A "303-3-303 ANN model" was developed with 303 neurons in input layer, three neurons in hidden layer and 303 neurons in output layers. The input variables were the left lower limb extension / flexion joint angles, and the output variables were left lower limb extension / flexion joint moments. The results revealed that the ANN model fitted the experimental data well indicating that the model developed in this study was feasible in the assessment of joint moments for CMJ.

KEY WORDS: artificial neural network, joint moment, CMJ.

INTRODUCTION: Vertical jump has been often used to assess an athlete's lower limb muscular strength and power (Bosco, 1999). However, the information of jumping height is not enough for athletes to promote the performance because it can only indicate the result of the action of muscles at the lower limbs to accelerate the body segments upward. Human limb motions are caused and controlled by joint moments (Zernicke, 1996). Therefore, joint moments can provide direct information on both the muscle strength and neuromuscular control.

Inverse dynamics analysis is needed to calculate joint moments, which requires not only kinematic data but also ground reaction forces (GRF) and anthropometric data (Robertson et al., 2004). Since the procedure of inverse dynamics analysis is relatively complicated, this is not a practical tool for on-field instructors and physical educators. It will be helpful to develop a quicker and simpler processing method in order to obtain the joint moments. Artificial neural network (ANN) is a computational technique that has the inclination for storing experimental knowledge and making it available for application (Schollhorn, 2004). The ANN modeling has been widely adopted in the area of clinical biomechanics. Liu et al. (2009) develop an ANN model for prediction joint moments at hip, knee and ankle joints using GRF during counter-movement jump (CMJ) and squat jump (SJ). The purpose of this study was to develop an ANN model for predicting lower limb joint moments at hip, knee and ankle using solely the relevant kinematic parameters during CMJ.

METHOD: Nine female students from a college volleyball team (age: 20±1.01 years, height: 165±6.24 cm, mass: 59±4.82 kg) performed CMJ three times (one random trial data of each subject was used). The kinematic and kinetic data were measured with a 7-camera motion analysis system (Vicon 512, Oxford Metrics, U.K.) and two force plates (AMTI, Advanced Mechanical Technology, U.S.A.). The joint moments at hip, knee and ankle were calculated by inverse dynamics (called "measured joint moments" in this study).

A feed-forward back propagation network comprised of one input, one hidden and one output layer was developed. The input variables of the ANN model were left limb extension / flexion joint angles, and the output variables were left limb extension / flexion joint moments during the support phase of CMJ. Since the movement times were different among subjects, all input and output variables were normalized to 100% of support phase. Moreover, the input variables were scaled and the output variables were rescaled before and after running the ANN model. The data of eight randomly chosen subjects were used for training of the ANN model. The other one subject's data were used to verify the performance of the ANN model between the ANN-predicted and inverse dynamics calculated values. The ANN model was implemented using Matlab 7.3.0 (R2006b).

RESULTS: After trial-and-error procedure, a “303-3-303 ANN model” with the minimal root mean square error (RMSE) was developed with 303 neurons in input layer, three neurons in hidden layer and 303 neurons in output layers. The figure 1-3 shows that the ANN-predicted curves of joint moments fit the measured curves well. The correlation coefficients between the measured and ANN-predicted joint moments during the support phase of CMJ (Table 1) were all more than 0.94. The results revealed a high level of agreement between the ANN-predicted joint moments and those experimentally measured.

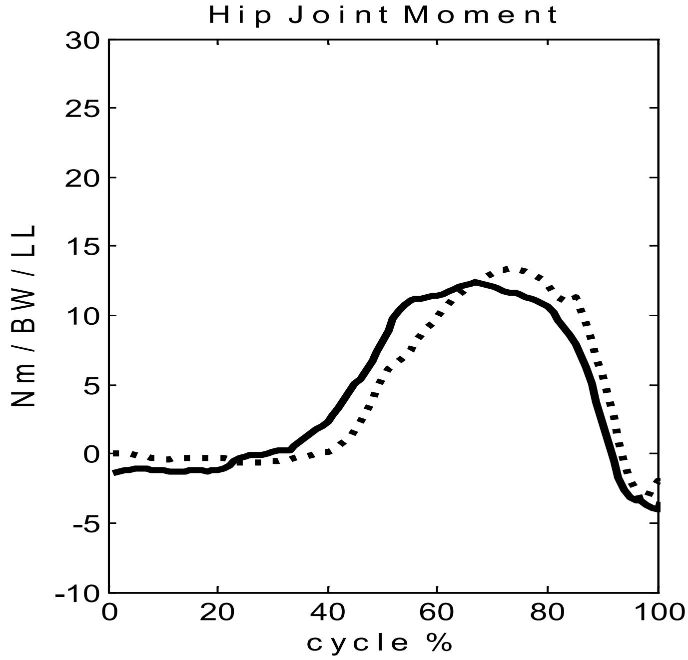

Figure 1. The measured (dotted line) and ANN-predicted (thick line) joint moments at hip during CMJ

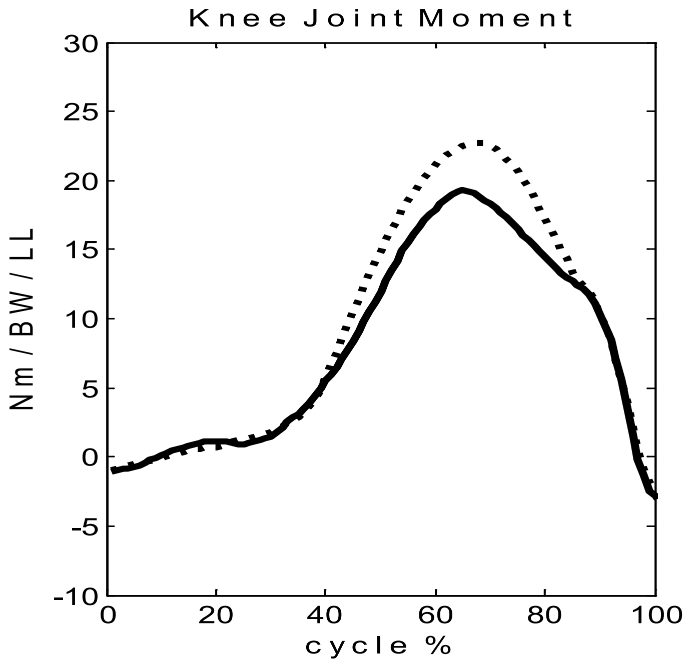

Figure 2. The measured (dotted line) and ANN-predicted (thick line) joint moments at knee during CMJ

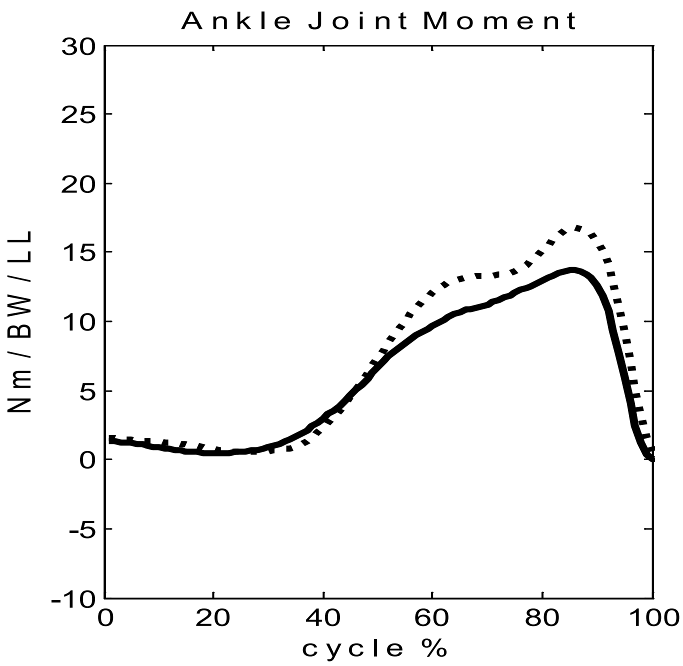

Figure 3. The measured (dotted line) and ANN-predicted (thick line) joint moments at ankle during CMJ

Table 1. The correlation coefficients (r) between the measured and ANN-predicted joint moments

Joint moments	r
Hip (extension / flexion)	0.9418
Knee (extension / flexion)	0.9958
Ankle (extension / flexion)	0.9920

DISCUSSION: The correlation coefficients between the measured and ANN-predicted joint moments during the support phase of CMJ were all more than 0.94, indicating a high level of agreement between the measured and ANN-predicted joint moments. The ANN-predicted and measured joint moments agreed very well both in trend and magnitude, suggested the ANN method with solely kinematics parameters is capable of producing very approximate measured joint moments of lower limbs during CMJ. In addition, through the fitting quality of the curves, it was confirmed that the ANN possesses the abilities of filtering and admitting noise (Schalkaff, 1997).
However, the results were from only one sport item, and there was only one subject's data used to verify the performance of network, so our ANN model must be used carefully for other kinds of population. In the future, training of the ANN-model with greater sample size of training data will increase its accuracy and will be helpful for its application in the evaluation of athletes' muscular strength and neuromuscular control of the lower limbs.

CONCLUSION: A feed-forward back propagation ANN model comprised of one input, one hidden and one output layers was developed by trial-and-error procedure to predict the lower limb joint moments using solely the relevant kinematic parameters during CMJ. The results indicate that the ANN model developed in this study is feasible in the assessment of lower limb joint moments without inverse dynamics calculation for CMJ. After improvement of the ANN model with large number of subjects, it is believed that this kind of ANN model might be used to evaluate in depth the athletes' joint moments in lower limbs.

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ISBS 2010

Oral Session 12

Kinetics and Kinematics

New Investigator Award

KINEMATICS OF THE TRUNK AND THE SPINE DURING UNRESTRICTED AND RESTRICTED SQUATS – A PRELIMINARY ANALYSIS

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The aim of the study was the assessment and comparison of trunk kinematics during restricted (knee not beyond toes) and unrestricted squats. Eight repetitions of restricted and unrestricted squats with 0, 25% and 50% bodyweight loading using a barbell were assessed with a 12 camera motion analysis system. A trunk marker set was developed and applied that allowed the measurement of the 3D kinematics of the trunk, divided into three segments (lumbar, thoracic and upper trunk) and the analysis of the sagittal curvature (lumbar and thoracic spine). The preliminary results of four subjects showed a larger range of sagittal motion between lumbar and pelvic segments for the restricted compared to the unrestricted squat. The lumbar curvature straightened with increasing load. The unrestricted execution seems to lead to higher stresses in the lower back.

KEYWORDS: squat, marker set, trunk, back, kinematics

INTRODUCTION: The squat exercise is one of the basics in fitness training, in strength training and in rehabilitation. In the bfu report 39 (bfu: Swiss Federal Office for Accident Prevention) the squat exercise was evaluated as one of the most predestinated exercises for injury-risk and complaint-risk (Müller, 1999). The squat can be performed in a restricted (**r**) and in an unrestricted (**unr**) type. In the **r** type the knees are only allowed to move until they reach the vertical line of the toes. The **r** type of the exercise is very often used in fitness centers (Chandler and Stone, 1991). However, it is unclear what effect the restriction of displacement in the lower limbs has on the kinematics of the trunk.
Most of the current trunk models based on skin marker assessment either consider the trunk as a single segment (Ferrarin et al., 2005; Kramers-de Quervain et al., 2004; Nguyen and Baker, 2004; Whittle and Levine, 1997) or describe spine motion (D'Amico et al., 1995; Frigo et al., 2003; Whittle and Levine, 1997). Crosbie et al. (1997) divided the trunk into three segments, namely lumbar, lower trunk and upper trunk, each defined by three skin markers allowing to describe three dimensional segmental kinematics. Compared to the huge range of different marker sets to assess the kinematics of the lower extremities, only very little work has been done concerning the trunk.
The kinematic investigation of the different types of the squat exercise is important for the strategy of appropriate strength training (Lorenzetti et al., 2009). Thus, the aim of the present study was to compare trunk motion between **r** and **unr** squats. This included the development of a suitable trunk marker set and its kinematic procedure to assess the kinematics of the trunk during squats based on skin markers.

METHOD: Twelve subjects, all movement science students experienced in weight lifting participated in this study. Four subjects were preliminary analysed. In average the four subjects weighed 67.5 ± 15.5 kg, showed a height of 175 ± 14 cm and an age of 24 ± 5 years.
The 3D motion analysis system used is a 12 camera VICON MX system (Oxford Metrics Group, UK). The used capture frequency was 50 Hz and the capture volume 300 cm x 500 cm x 200 cm. The used marker set for the assessment of trunk and pelvis kinematics consisting of 31 skin markers with a diameter of 9 and 14 mm is shown in Figure 1 and Table 1. The allocation of the markers to the used segments is shown in Table 2.
First, the subjects had to perform a standing trial in an anatomic upright position. Following, the subjects performed **r** and **unr** squats with zero, 25% bodyweight (BW) and 50% BW loading using a barbell. Each of the six conditions consisted of eight repetitions. For the **r** squat, the knee was not allowed to go beyond the toes. This restriction was visually self-controlled by the subject with the use of a live projection of the side view of knee and toes

and a pile marking the front edge of the toes onto a screen in front of the subject. No external force was applied to restrict the motion of the knee. The **unr** squat was performed with no restriction on the motion of the knee.
Two approaches were used to describe the kinematics of the trunk, a segmental and a curvature approach.
Segmental approach: The position and orientation of each segment was determined relative to the reference segments defined by the standing trial using a least-squares fit of the corresponding marker point clouds (Gander and Hrebicek, 1997). It follows that the neutral position (0° rotation) was defined by the standing trial. Each segment was defined by a redundant number of markers, aiming in an improvement in orientation accuracy (Challis, 1995). Joint rotations were described from the lower relative to the upper segment (pelvic relative to lumbar, lumbar relative to thoracic and thoracic relative to upper trunk segment) using a helical axis approach (Woltring, 1994). To define clinically interpretable rotational components the attitude vector was decomposed along the axes of a marker based joint coordinate system (Woltring, 1994). The vertical axis e_v of the joint coordinate system, connecting line between the spine markers L5 and C7, is the leading axis, pointing cranial. The transverse axis e_t is perpendicular to e_v and lies in the plane spanned by e_v and the connecting line between the markers RTBL and LTBL, pointing from left to right. The posteroanterior axis e_{pa} is perpendicular to the latter two and points to the front. Hence, clinical rotations are described as follows: rotation around e_v stands for axial rotation (positive rotation denotes a frontal motion of the left upper segment with respect to the lower), around e_t for flexion/extension (positive rotation corresponds to forward flexion) and around e_{pa} for lateral bending (positive rotation stands for bending to the left side).
For the assessment of the sagittal plane curvature of the lumbar and the thoracic spine, the corresponding marker positions were projected on to the sagittal trunk plane defined by the plane spanned by e_v and e_t. The curvature was estimated by the reciprocal of the radius of the circle that was fitted into the corresponding five markers using a least-squares approach.
A squat cycle was defined as starting in a more or less upright position, moving down to the lowest position and up again. The start and end point of the cycle was defined by the vertical velocity of the barbell (v_{barb} > 0.02m/s). For each condition mean and standard deviation (SD) over the eight repeated cycles were calculated.
ROM was defined as the range between minimal and maximal reached rotation values.

Table 1: Marker placement and abbreviations.

RTSH, LTSH	right and left acromion
RTCL, LTCL	right and left clavicle
STER	sternum
RTSC, LTSC	right and left inferior angle of the scapula
RTBH, LTBH	right and left most inferior rib
RTBL, LTBL	right and left lateral back on height of L4
C3 , C5, C7	third, fifth and seventh cervical vertebrae
T3, T5, T7, T9, T11	third, fifth, seventh, ninth and eleventh thoracic vertebrae
L1, L2, L3, L4, L5	first, second, third, fourth and fifth lumbar vertebrae
RTAS, LTAS	right and left anterior superior iliac spine
RTPS, LTPS	right and left posterior superior iliac spine
RTMS, LTMS	right and left mid superior iliac spine
SACR	sacrum

Table 2: Segmental allocation.

Lumbar spine	L1, L2, L3, L4, L5
Thoracic spine	T3, T5, T7, T9, T11
Pelvic segment	RTAS, LTAS, RTPS, LTPS, RTMS, LTMS, SACR
Lumbar segment	L1, L2, L3, L4, L5, LTBL, RTBL
Thoracic segment	T3, T5, T7, T9, T11, LTSC, RTSC, LTBH, RTBH, STER
Upper trunk segment	C7, LTSH, RTSH, RTCL, LTCL

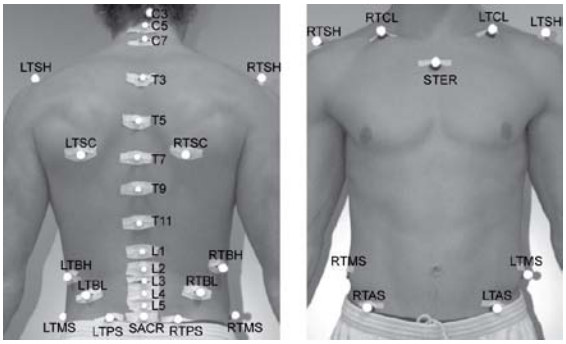

Figure 1. Marker set of the trunk and the pelvis. For explanation of abbreviations see Table 1, for segmental allocation see Table 2.

RESULTS: For all weight conditions and squat types segmental rotation was predominant in the sagittal plane, rotation in the frontal and the transverse plane were small (mean over four subjects: ROM lateral bending < 2.8°, ROM axial rotation < 3.6°). Pelvic relative to lumbar segmental sagittal rotation showed larger ROM for **r** (mean over four subjects: 0% BW: 22.7 ± 3.2°, 25% BW: 20.9 ± 2.7°, 50% BW: 18.3 ± 2.5°) compared to **unr** (mean over four subjects: 0% BW: 19.1 ± 2.6°, 25% BW: 17.9 ± 4.6°, 50% BW: 17.3 ± 1.8°) (see Figure 2). The lumbar spine curvature decreases with additional load, whereas the thoracic curvature is not dependent on loading (see Figure 3).

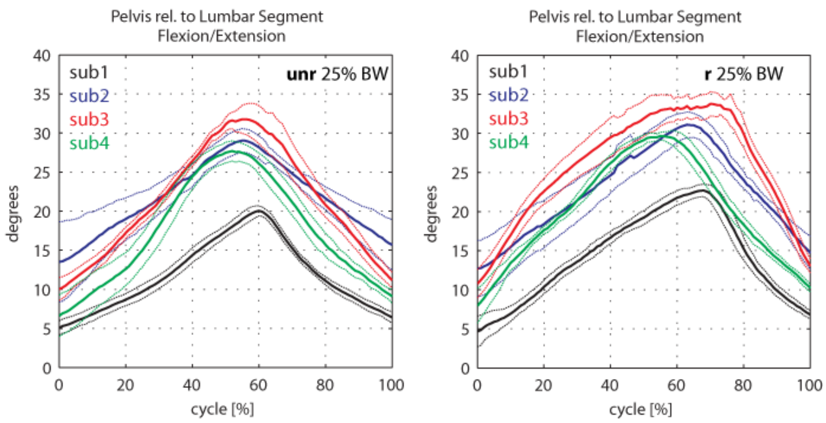

Figure 2. Pelvic relative to lumbar segmental sagittal rotation. Mean and SD over 8 cycles. unr 25% BW (left), r 25% BW (right).

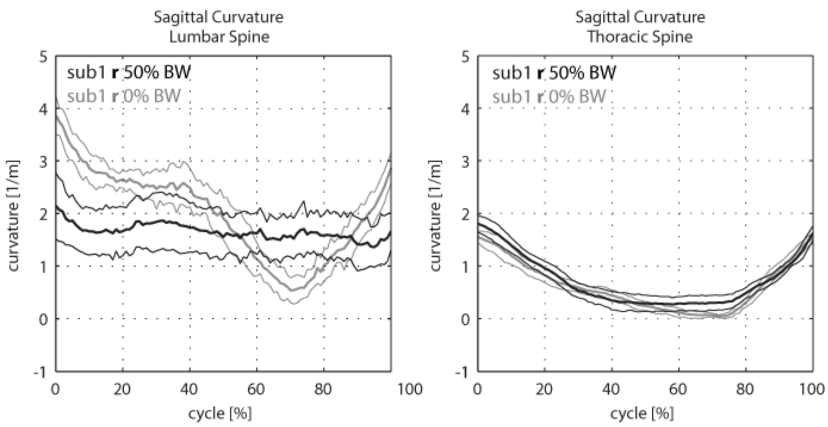

Figure 3. Sagittal curvature of the lumbar (left) and thoracic (right) spine of sub1. Mean and SD over 8 cycles. r 0% BW (grey), r 50% BW (black).

DISCUSSION: A marker set and its corresponding data processing has been developed that allows the assessment of the kinematics of the trunk in terms of a 3D segmental approach based on three trunk and one pelvic segment as well as in terms of a sagittal plane spine curvature analysis. Both approaches are suitable to assess the movement of the trunk during squatting.
A restriction in knee motion results in an increased trunk flexion. Assuming a simple mechanical model, this leads to higher stresses in the lower back. The load dependent straightening of the spine is in agreement with the study of Meakin et al. (2008).

CONCLUSION: In this study the 3D segmental motion of the trunk, based on a three segment trunk model, as well as the spinal sagittal curvature was determined. Given by the skin marker approach, the present method is limited by skin movement artefacts.
Not surprisingly, the ROM of flexion between the pelvic and the lumbar segment during squatting increases with a restriction in knee motion. Therefore, the stress on the lower back most likely is lower during an unrestricted squat. For these reasons, the unrestricted squat may be the right choice for most athletes.

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KINETIC ANALYSIS OF SEVERAL VARIATIONS OF PUSH-UPS

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Push-ups are a common and practical exercise though the kinetic characteristics of this exercise and its variations have yet to be quantified. This study assessed the peak ground reaction forces (GRF) of push-up variations including the regular push-up and those performed with bent knee, feet elevated on a 30.48 cm box and a 60.96 cm box, hands elevated on a 30.48 cm box and a 60.96 cm box. Peak GRF and peak GRF expressed as a coefficient of subject body mass were obtained with a force platform. Push-ups with the feet elevated produced higher GRF than all other push-up variations ($p \leq 0.05$). Push-ups with hands elevated and from the bent knee position produced lower GRF than all other push-up variations ($p \leq 0.05$). These data can be used to progress the intensity of push-ups in a program with loads that are quantified as a percentage of body mass.

KEYWORDS: strength, body weight, upper body, closed kinetic chain

INTRODUCTION: Push-ups are a commonly performed, practical, easy to execute, multi-joint upper body exercise that do not require expensive equipment. Push-ups have been recommended as one of the best practical upper body exercises in a popular consumer publication (Lee, 2008). Push-ups are one of a limited number of closed kinetic upper body exercises (Blackard et al., 1999) and there are many potential variations (Osbourne, 1989). Unfortunately, the quantification of this exercise stimulus in a training program is difficult compared to more traditional resistance training exercises. Resistance training exercises are often performed with equipment such as a barbell and weight plates with clearly labeled masses. These loads are often calculated and exercises are performed with a percentage of the exercisers maximum ability. Determination of the intensity of a resistance training stimulus allows for the progression of exercise intensity and the calculation of training volume.

Push-ups have also been evaluated as an upper body strength test (Mayhew et al., 1991) and have been demonstrated to increase upper body strength and power (Vossen et al., 2000). Research quantifying push-ups and the variations of this exercise is limited. The traditional push-up has been compared to a manufactured product purported to increase the quality of the push–up training stimulus using electromyography and video analysis to assess the differences (Bohne, et al., 2009). Research using electromyography to assess muscle activation demonstrated that variations in hand stance width resulted in different levels of muscle activation of the pectoralis major and triceps brachii (Cogley et al., 2005). Only one study used ground reaction forces (GRF), as well as electromyography, to assess variations in push-ups characterized by differences in hand position as well as a bent knee condition. This study described the push-up as a percentage of body weight and demonstrated differences in GRF between variations of hand placement and between bent knee and normal push-up conditions (Gouvali & Boudolos, 2005). Many variations of push-ups have yet to be studied including the conditions where feet are elevated or where hands are elevated. The purpose of this study was to assess the GRF associated with regular push-ups, and those performed with bent knee, feet elevated on a 12” box, feet elevated on 24” box, hands elevated on a 12” box, and hands elevated on a 24” box for the purpose of quantifying the intensity of these exercises for exercise progression and to allow for the calculation of exercise load and volume in a program. This study also sought to assess if there were gender based differences in response to these push-up variations.

METHODS: Twenty-three recreationally fit young adults (mean ± SD; age = 22.5 ± 4.6 years, height = 178.10 ± 10.89 cm; body mass = 80.59 ± 9.28 kg) volunteered to serve as subjects for the study. Subjects signed an informed consent form and Institutional Review Board approval was obtained prior to the study.

Subjects were instructed in and received demonstration of each push-up variation including a regular push-up, and those performed with bent knee, feet elevated on a 30.48 cm box, feet elevated on 60.96 cm box, hands elevated on a 30.48 cm box, and hands elevated on a 60.96 cm box. For all warm up, practice and test push-ups, subjects hand placement was defined as the width equal to the distance of contralateral acromion processes measured from the inside border of each hand with hands placed under the shoulders in the beginning position which was characterized by full elbow extension.

Subjects warmed-up by performing 3 practice repetitions of each push-up variation in a randomized order. Subjects then rested for 2 minutes. Subjects then performed 2 repetitions of each push-up variation in randomized order. Subjects rested for 1minute in between each push-up variation. A metronome was used to control the cadence of the push-up repetitions with each repetition performed for a count of two seconds. A fast pace was desired since previous research demonstrated higher levels of power and work for fast compared to slower cadences (LaChance and Hortobgyi,1994) and the 2 second cadence has been demonstrated to be effective at increasing upper body strength (Vossen, et al., 2000).

The push-up variations were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies, Inc., Watertown, MA, USA), which was calibrated with known loads to the voltage recorded prior to the testing session. Data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA, USA) for later analysis. All values were determined as the average of two trials for each push-up variation. Peak GRF and peak GRF expressed as percentage of static body mass GRF were evaluated from the vertical force-time records using custom designed software. Data were analyzed using SPSS 16.0 with a repeated measures ANOVA to determine possible differences between push-up variations. Bonferroni adjusted pairwise comparisons identified the specific differences between these variations. The *a priori* alpha level was set at $p \leq 0.05$. Effect sizes and power are reported as η_p^2 and *d*, respectively.

RESULTS: There was a significant main effects for push-up condition for body weight coefficient ($p \leq 0.001$, $\eta_p^2 = 0.94$, $d = 1.00$), with no interaction between push-up condition and gender for this variable ($p = 0.25$). There was a significant main effect for peak GRF ($p \leq 0.001$, $\eta_p^2 = 0.76$, $d = 1.00$) with no interaction between condition and gender for this variable ($p \leq 0.05$). Trial to trial reliability of the body weight coefficient were moderately to highly reliable as demonstrated by Intraclass Correlation Coefficient values in a range from 0.64 to 0.84 with no significant differences between trials ($p > 0.05$). Trial to trial reliability of the mean peak GRF were highly reliable as demonstrated by Intraclass Correlation Coefficient values in range from 0.97 to 0.99 with no significant differences between trials ($p > 0.05$).

Table 1. Push up peak ground reaction force expressed as a percentage of subject static body mass peak ground reaction force.

Push-Up Variation	Body Weight Coefficient
Feet Elevated 60.96 cm	0.74 ± 0.02*
Feet Elevated 30.48 cm	0.70 ± 0.02*
Regular	0.64 ± 0.04*
Hands Elevated 30.48 cm	0.55 ± 0.05*
Bent Knee	0.49 ± 0.05*
Hands Elevated 60.96 cm	0.41 ± 0.06*

*Significantly different than all other push up conditions ($p \leq 0.01$)

Table 2. Push up peak ground reaction expressed in Newtons (N) for each push up variation.

Push-Up Variation	Peak GRF
Feet Elevated 60.96 cm	723.89 ± 138.24¥
Feet Elevated 30.48 cm	714.05 ± 142.61¶
Regular	700.13 ± 138.25*
Hands Elevated 30.48 cm	681.46 ± 125.31*
Hands Elevated 60.96 cm	596.75 ± 113.43*
Bent Knee	569.92 ± 138.24*

*Significantly different ($p \leq 0.05$) than all other push up conditions.
¶ Significantly different ($p \leq 0.05$) than all other push up conditions except Feet Elevated 60.96 cm
¥ Significantly different ($p \leq 0.05$) than all other push up conditions except Feet Elevated 30.48 cm

DISCUSSION: This is the first study to assess push-up variations that include varying levels of feet and hand elevation, demonstrating differences between these conditions. Results of this study can be used to guide the progression of overload, by incorporating the push-up variations with higher GRF's over time. Progression of overload is believed to be important for exercise program design (Fleck & Kraemer, 1997).
Results of this study demonstrated GRFs of approximately 64% and 49% of body weight in the regular and bent knee push-up conditions, respectively. These findings were slightly different than the values of 66% and 53%, for the regular and bent knee push up variations, respectively, previously found (Gouvali & Boudolos, 2005). Changes in hand placement produce GRFs in a range from 52.9% to 72.9% of body mass (Gouvali & Boudolos, 2005), whereas the present study demonstrates that push-up conditions that included hand and feet elevation produced values that ranged from approximately 41% to 74% of body mass.

CONCLUSION: Practitioners should progress push-up intensity from lower intensity push-up variations such as the elevated hand and bent knee conditions to normal push-ups to the feet elevated conditions. These data can also be used to quantify the approximate load, as a percentage of body mass for the purpose of quantifying exerciser load and volume in a resistance training program.

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THE KINEMATICS OF TETHERED WALKING AND JOGGING

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Twelve subjects (11 male, 1 female) volunteered to walk and jog on a treadmill during tethered and no tethered conditions. The four randomly applied conditions were no tether (control), arms tethered (AO), legs tethered (LO), arms and legs tethered (AL). Three consecutive gait cycles were recorded and digitized to obtain trunk angle, thigh angular displacement and velocity, arm angular displacement and velocity, and vertical displacement of center of gravity. Significant differences were found between the kinematic variables in tethered and no tethered conditions ($p \leq 0.05$). Further analysis is needed to determine the effect of tethering on stride length and stride rate. A follow-up training study is planned to determine whether tethered training conditions improve a runner's speed.

KEYWORDS: angular kinematics, resisted locomotion, training techniques, elastic cords

INTRODUCTION: Speed, power, and agility provide a solid foundation upon which an athlete can build his/her repertoire of sport-specific skills. Dillman (1975) and Williams (1985) have suggested that running speed is a product of the athlete's stride length and stride rate. Increases in running speed are made as stride length is discretely increased from 3.5 up to 6.5 m/sec however, for higher speeds, stride rate must be increased (Dillman, 1975; Williams, 1985).
Because running speed is an essential component in many sporting events, runners and coaches utilize various training techniques that will augment a runner's speed. Various resistance training methods such as sled towing, partner tethering, parachute towing, limb and trunk loading, and elastic cord tethering to stationary objects have been utilized to improve running speed. Several researchers have investigated the effects of these various training techniques on the runners' kinematics and have reported mixed results (Martin, 1985; Letzelter et al., 1995; Lockie, Murphy, & Spinks, 2003; Corn & Knudson, 2003). Martin (1985) added various loads to the thighs and feet of 15 male runners and reported small, but significant increases in stride length only when loads of 0.5 kg were added to the feet. Letzelter et al. (1995) analyzed the kinematics of 16 trained female sprinters using resisted sled towing. Letzelter and colleagues (1995) reported decreases in stride length and stride rate as well as increases in trunk lean. Corn and Knudson (2003) reported significant increases in the stride length and horizontal velocity of the center of mass in collegiate sprinters when tested in partner tethered sprinting. Lockie et al. (2003) found significant decreases in stride length when sprinters towed sleds weighted with loads equivalent to 12.6% and 32.2% of the subject's body mass. The results of these previous studies have been mixed, with some techniques causing increases and other techniques causing decreases in the runner's kinematics. The purpose of this study was to investigate the kinematic responses to tethering the subjects' arms, legs, and arms and legs to a stationary object while the subject walked and jogged on a treadmill. These conditions were selected because they have been utilized by speed development specialists. Due to safety concerns regarding the tethering of the subjects' extremities, slower forms of locomotion, i.e., walking and jogging were used in this study in place of sprinting.

METHODS: Twelve, healthy subjects (11 male, 1 female) gave written consent in compliance with Institutional Human Subjects Review Committee Guidelines. Subjects were required to complete a treadmill familiarization protocol in which each subject walked at the preselected speeds of 4.83 km/hr and jogged at 9.68 km/hr for 15 minutes at each speed. Subjects returned on a second day to have their cardiovascular fitness measured and to determine a baseline VO_2max using the Bruce treadmill test protocol. Subjects were given a

72 hour rest period following the Bruce test. Kinematic data collection required two days to complete, with a 48 hour rest given between the two sessions. The subjects were outfitted in black compression shorts and shirts. Reflective markers were fixed to the following anatomical landmarks: mastoid process, lateral tubercle of humerus, lateral epicondyle of humerus, greater trochanter, lateral epicondyle of femur, lateral malleolus, and fifth metatarsal head. A digital video camera operating at 60 Hz was positioned 8 m from the midline of the treadmill and perpendicular to the sagittal plane of motion. Three consecutive gait cycles were recorded for each condition as subjects walked at 4.83 km/hr and jogged at 9.68 km/hr for five minutes per condition.
The four randomly applied conditions were no tether (control), arms tethered (AO), legs tethered (LO), and arms and legs tethered (AL). Bilateral tethering was achieved by fixing one end of the Power sprinter stretch cords to a nylon cuff that was placed at mid thigh and/or mid humerus. The other end of the stretch cord was attached to a force transducer (Vernier, dual-range force sensor) that was fixed to a stationary vertical pole. Data were digitized and analyzed using the Ariel Performance Analysis System (APAS). Trunk angle, thigh angular displacement and velocity, upper arm angular displacement and velocity, and vertical displacement of the center of gravity were measured. The stretch cord forces were recorded throughout the tethered conditions. A two way repeated measures ANOVA and Holm-Sidak post hoc test were used to statistically determine differences in the angular kinematics between condition and speed.

RESULTS AND DISCUSSION: The kinematic data for walking are displayed in Table 1. The angle of trunk inclination in the NT condition was significantly different from the tethered conditions during walking. The thigh angular displacement in the NT condition was also significantly different than the tethered conditions. Also LO was significantly different from the AL and AO conditions. For thigh angular velocity, the NT condition was significantly different from AL and AO only. Arm angular displacement showed significant differences between the NT and the AL and AO conditions, with the NT condition going through a greater range of motion. There were no significant differences noted for arm angular velocity or vertical displacement of the COG during walking. Throughout the data collection, stretch cord tension was monitored and recorded. The stretch cord tension for the AL condition was significantly different from the AO and LO conditions. Subjects self-reported that the AL condition was more difficult than the other conditions.

Table 1. Means $\pm$ SD at walking speed of 4.83 km/hr.

Variable	Arms & Legs Tethered	Arms Tethered	Legs Tethered	No Tether
Trunk Angle (degrees)	87.4 $\pm$ 4.2	88.7 $\pm$ 3.4	88.5 $\pm$ 3.3	92.9 $\pm$ 1.0*
Thigh Displacement (degrees)	35.5 $\pm$ 2.8	34.7 $\pm$ 2.7	30.9 $\pm$ 1.5*	25.9 $\pm$ 1.2*
Thigh Velocity (degrees/sec)	51.1 $\pm$ 5.5	51.6 $\pm$ 5.0	45.7 $\pm$ 2.9	38.5 $\pm$ 2.6*
Arm Displacement (degrees)	16.8 $\pm$ 3.5	18.4 $\pm$ 2.9	22.9 $\pm$ 2.4	25.8 $\pm$ 1.7*
Arm Velocity (degrees/sec)	31.8 $\pm$ 1.9	45.8 $\pm$ 1.5	42.5 $\pm$ 1.2	49.9 $\pm$ 1.1
COG Displacement (cm)	1.8 $\pm$ 0.3	1.9 $\pm$ 0.4	1.7 $\pm$ 0.2	1.5 $\pm$ 0.3
Cord Tension (N)	16.8*	8.0	8.8	

* Statistically significant at $p \leq 0.05$

The kinematic data for jogging are displayed in Table 2. During jogging, the AL and AO conditions were significantly different from the LO and the NT conditions. The thigh angular displacement in the NT condition was significantly different from AL and AO, but not from LO. Likewise LO was significantly different from AL and AO. Thigh angular velocity for AL and AO was significantly different from the LO and NT conditions. For arm displacement, the tethered arm condition, AO, was found to be significantly different from the LO condition only. These results are consistent with what one would expect when the arms are tethered to a stationary object. Unlike the walking condition, a significant difference in arm angular velocity was observed between AL and the NT and AO conditions. There were no differences between the AL and LO conditions for arm angular velocity. Similar to the walking condition, no significant differences were noted for the vertical displacement of the COG. The elastic cord tension for the AL condition was significantly different from the AO and LO conditions. Subjects reported feeling exhaustion similar to what they experienced during the Bruce treadmill test protocol.

Table 2. Means ± SD at jogging speed of 9.68 km/hr.

Variable	Arms & Legs Tethered	Arms Tethered	Legs Tethered	No Tether
Trunk Angle (degrees)	78.7 ± 5.3*	78.9 ± 5.2*	85.1 ± 4.9	85.7 ± 4.7
Thigh Displacement (degrees)	44.9 ± 5.7	42.0 ± 4.8	38.9 ± 3.4*	35.9 ± 2.5*
Thigh Velocity (degrees/sec)	159.9 ± 5.2*	149.2 ± 3.5*	137.4 ± 3.4	133.1 ± 1.8
Arm Displacement (degrees)	36.2 ± 2.9	31.8 ± 2.3*	39.4 ± 1.7	33.4 ± 1.2
Arm Velocity (degrees/sec)	168.7 ± 3.2*	143.3 ± 2.7	150.3 ± 2.0	138.9 ± 1.2
COG Displacement (cm)	7.5 ± 1.1	7.3 ± 1.0	7.3 ± 1.2	7.6 ± 1.1
Cord Tension (N)	17.4*	8.8	9.0	

* Statistically significant at $p \leq 0.05$

At both the walking and jogging speeds, the thigh angular displacement and angular velocity were greatest for the AL condition. According to Dillman (1975) and Williams (1985), if an athlete wanted to increase running velocity, stride length and stride rate would have to increase. Although stride length and stride rate were not measured during this phase of the study, the increase in both thigh angular displacement and angular velocity during the AL condition may result in an increase in both stride length and stride rate. If resisted jogging were performed with enough frequency, this training effect might carry over into the athlete's normal running pattern. It is interesting to note that the same pattern was not observed for the upper arm angular displacement or velocity. During walking, the AL condition resulted in the smallest angular displacement and angular velocity of the arm; however, the same was not observed during the jogging speed. The AL condition accounted for the second largest angular displacement of the arm and the highest angular velocity.

CONCLUSION: Results of this study demonstrate that tethered walking and jogging can alter the angular kinematics. The biomechanical responses to the tethering do not appear to be in opposition to what an athlete would want to achieve through training; namely, increased stride length and stride rate which translates into an increased horizontal velocity.

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A COMPARISON OF PRE- AND POST-OPERATIVE THREE-DIMENSIONAL HIP KINEMATICS DURING LEVEL WALKING IN PATIENTS WITH CAM FEMOROACETABULAR IMPINGEMENT

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KEYWORDS: femoroacetabular impingement (FAI), hip kinematics, level walking.

INTRODUCTION: Cam femoroacetabular impingement (FAI) is an idiopathic progressive pathological condition of the hip joint characterized by an abnormal bony protuberance on the femoral head-neck junction (Beck, Leunig, Parvizi, Boutier, Wyss & Ganz, 2004). During the limits of hip range of motion (ROM), the protuberance jams into the acetabulum (Ganz, Parvizi, Beck, Leunig, Nötzli & Siebenrock, 2003), resulting in acute hip and groin pain (Beaulé, LeDuff, & Zaragoza, 2007). Impingement has also been shown to occur within normal ROM of the hip during basic tasks such as walking, reducing peak hip abduction angles as well as hip frontal and sagittal ROM (Kennedy, Lamontagne & Beaulé, 2009). Cam FAI primarily affects young and athletic males (Ganz, Parvizi, Beck, Leunig, Nötzli & Siebenrock, 2003), and is common in hockey, football, soccer, rugby, martial arts and tennis athletes (Philippon, Schenker, Briggs & Kuppersmith, 2007). Restricted hip mobility during activities requiring low ROM suggests more pronounced limitations during demanding athletic tasks. Surgical procedures have been developed to remove the bony abnormality from the femoral head-neck junction with the objective of attenuating hip pain and restoring normal hip biomechanics, enabling athletes to return to sport. The purpose of this study is to assess the clinical outcome of cam FAI corrective surgery by comapring pre-operative and post-operative three-dimensional (3-D) hip kinematics during level walking.

METHOD: Five participants (4 males, 1 female) took part in the pilot study. Each participant was tested pre-operatively and post-operatively. Posttests were performed a minimum of 11 months following surgical intervention. All surgeries were performed by the same surgeon and involved the dislocation of the hip and debridement of the femoral head-neck junction.
An infrared nine-camera high-speed motion analysis system (Vicon MX-13, Oxford Metrics, Oxford, UK) was used to capture the 3-D kinematics at 200 Hz. Participants wore form-fitting spandex shirt and shorts to reduce clothing artefact during motion trials. Retro-reflective markers were affixed to the participants according to a modified version of the Plug-in Gait marker set (Vicon, Oxford Metrics, Oxford, UK). First, a static calibration of the participants was performed to calculate the segment lengths and determine neutral joint positions. Then, participants performed five barefoot level walking trials at a self-selected pace.
All five trials were averaged for each participant and then ensemble-averaged. A series of repeated measures analysis of variance (ANOVA) were conducted to compare pre- and post-operative hip kinematics values. Specifically, hip ROM in each plane as well as peak angular displacements in flexion/extension, abduction/adduction and internal/external rotation were assessed. Since multiple variables were compared, a Bonferroni adjustment was made for the α value. Therefore, the significance level for all statistical analyses was set at $\alpha \leq 0.0167$.

RESULTS: Hip angular displacements in the sagittal and frontal planes during level walking are displayed in Table 1. Statistical analyses revealed no significant differences between any of the pre-operative and post-operative hip kinematics values of interest, with p values ranging from 0.274 to 0.916.

Table 1. Hip Angular Displacements during Level Walking

Participant	Time	Peak Flexion	Peak Extension	Sagittal ROM	Peak Adduction	Peak Abduction	Frontal ROM
1	Pre	39.14°	-9.50°	48.63°	13.75°	-0.44°	14.20°
	Post	36.31°	-14.54°	50.85°	11.64°	-1.76°	13.40°
2	Pre	27.58°	-14.37°	41.93°	14.70°	1.44°	13.26°
	Post	27.03°	-11.53°	38.57°	11.29°	-1.14°	12.42°
3	Pre	30.58°	-16.86°	47.45°	5.84°	-7.50°	13.34°
	Post	27.17°	-17.18°	44.35°	7.23°	-5.33°	12.56°
4	Pre	31.14°	-17.82°	48.96°	6.44°	-6.42°	12.86°
	Post	34.09°	-19.44°	53.53°	11.25°	-7.13°	18.38°
5	Pre	30.01°	-13.31°	43.32°	11.19°	-5.02°	16.20°
	Post	34.73°	-10.60°	45.32°	11.32°	-9.97°	21.29°
Average	Pre	31.69°	-14.37°	46.06°	10.38°	-3.59°	13.97°
	Post	31.87°	-14.66°	46.52°	10.55°	-5.07°	15.61°
Standard Deviation	Pre	±4.38°	±3.28°	±3.22°	±4.09°	±3.89°	±1.34°
	Post	±4.42°	±3.72°	±5.86°	±1.86°	±3.70°	±4.01°

DISCUSSION: The objective of this study was to compare pre- and post-operative hip angular displacements during level walking, assessing the outcome of FAI cam impingement surgery. Kennedy, Lamontagne and Beaulé (2009) compared hip angular displacements during level walking between a group of 17 patients with cam FAI and a group of 14 healthy controls matched for age, sex and body mass index. The FAI group was found to have a lower peak hip abduction angle, a reduced hip frontal ROM and a slightly lower hip sagittal ROM during gait at a self-selected speed. These results suggest that corrective surgery for cam FAI should aim to restore these biomechanical parameters to within the normal range of values. Nevertheless, preliminary data from the current study show that peak hip angular displacements as well as hip ROM in each plane are not significantly different from those preceding surgery.

CONCLUSION: Pilot data indicate that corrective surgery for cam FAI does not restore hip kinematics to within a normal range of values during gait. Consequently, these results, perhaps transferable to more demanding athletic tasks, advocate that cam FAI debridement may not help athletes in returning to their normal sports practices. However, a larger sample is needed to solidify these findings, and more demanding activities need to be analyzed.

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INDIVIDUALIZED OPTIMAL RELEASE ANGLE IN DISCUS THROWING

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KEY WORDS: discus throwing, biomechanics, angle of release

INTRODUCTION: Release angle is a parameter that significantly influences the official distance through its effects on the vacuum flight distance and the aerodynamic distance (distance gain or lose due to aerodynamic effect) in discus throwing. Significant efforts have been made to determine the optimal release angle for discus throwing. Previous studies on the optimal release angle in discus throwing were based on two critical assumptions: (1) the release angles and the release speeds were independent of each other, and (2) the optimal release angle was the same for all discus throwers. These two assumptions may have been violated in previous studies on the optimal release angle (Hubbard et al., 2001; Linthorne, 2001; Viitasalo et al., 2007). The violation of these two assumptions may have resulted in significant errors in the optimal release angles for discus throwers reported in the current literature. The purposes of this study were: (1) to determine the relationships between the release speed and the release angle for individual discus throwers, (2) to determine the relationships between the aerodynamic distance and the release angle for individual discus throwers, and (3) to determine the optimal release angles for individual discus throwers; using data collected in competition.

METHOD: Three male and three female right-handed elite discus throwers were used as the subjects for this study. Each of these six subjects had at least 10 legal trials videotaped during the men's and women's discus throw competitions of the USA Track & Field National Outdoor Championships from 1997 to 2006. Two video camcorders were used to record the discus throwers' performances at a frame rate of 60 fields per second and a shutter speed of 1/1000 seconds using a setup for Direct Linear Transformation (DLT) procedure. Real-life three-dimensional (3-D) coordinates of the centre of the discus were estimated using the DLT procedure. The release speed and release angle of the discus, and the vacuum flight distance, the aerodynamic distance, and the distance lost at release were estimated from the 3-D coordinates of the discus. The original data set of each subject was randomly re-sampled to form 10 new data sets using a Bootstrapping method. The number of samples in each new data set was the same as the number of trials in the original data set. Multiple regression analyses were performed for each data set to express the release speed of the discus and the aerodynamic distance as a function of the release angle. The official distance of each subject was then expressed as the sum of vacuum and aerodynamic distances which were expressed as a function of the release angle using the regression relationships between the release speed and angle and between aerodynamic distance and release angle. Official distances were determined for each data set with the release angle varied from 30 to 50 degrees. The release angle corresponding to the longest official distance was considered as the optimal release angle for a given data set. The mean optimal angle of the 10 new data sets was considered as the optimal angle for a given subject. The 95% confidence interval around the mean optimal release angle for the 10 data sets for a given subject was used as the uncertainty measure of the estimated optimal release angle for a given subject.

RESULTS: The relationship between the release speed and the release angle was linear for all subjects. The regression coefficients in the equations relating the release speed to the release angle were different among subjects. The relationship between the aerodynamic distance and the release angle was also linear for all subjects. The regression coefficients in the regression equations relating the aerodynamic distance to the release angle were different among subjects.

Each subject had a single estimated optimal release angle corresponding to the longest calculated official distance (Table 1). The estimated optimal release angles were different among subjects and ranged from 35 to 44 degrees with uncertainties within 2 degrees (Table 1).

Table 1. Observed and Optimal Release Angles (degrees).

		Observed			
Subject	Gender	Minimum	Mean ± SD	Maximum	Optimal ± 95%CI
A	M	31	36 ± 2	38	40 ± 1
B	M	34	38 ± 2	41	44 ± 2
C	M	33	39 ± 3	41	39 ± 1
D	F	35	38 ± 2	43	44 ± 1
E	F	30	37 ± 3	44	35 ± 1
F	F	32	36 ± 3	44	37 ± 1

DISCUSSION: The relationships between release speed and angle of individual discus throwers observed in this study are consistent with those for javelin throwers (Viitasalo et al., 2007) and shot putters (Hubbard et al., 2001; Linthorne, 2001). These relationships indicate that the throwers' abilities to generate the speed of the implement change as the release angle changes.

The release angles preferred by elite discus throwers may not necessarily be optimal for themselves, and certainly not for all discus throwers. The optimal release angle is different for different discus throwers. The optimal release angles for individual discus throwers could be beyond the observed range of the release angles they currently used. Further wind tunnel studies are needed to determine the aerodynamic effects on optimal release angles with the consideration of the relationship between release speed and release angle. Future studies are also needed to determine the factors that affect the relationship between release speed and release angle. These studies will provide significant information for the technical and physical training of discus throwing.

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EFFECTS OF MOVEMENT SEQUENCE ON THE PERFORMANCE OF JAVELIN THROWING

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KEYWORDS: javelin throwing, movement sequence, biomechanics

INTRODUCTION: The sequence of joint and segment movements during delivery phase is believed to significantly affect on the performance of javelin throwing. Literature reported that elite javelin throwers showed a proximal to distal sequence of joint center maximum linear velocities (Whiting et al., 1991; Best et al., 1993; Mero et al., 1994; Bartlett et al., 1996). The sequence of maximum joint center linear velocities, however, does not necessarily represent the sequence of joint or segment movements. The purpose of this study is to determine the effects of the sequence of upper extremity joint and segment angular movements on the performance of javelin throwing.

METHOD: The subjects of this study were 32 male and 30 female right handed elite javelin throwers who competed in the 2007 and 2008 USA Track and Field Outdoor National Championships. The trial with the longest official distance for each thrower was used for this study. Subjects in each gender were divided into a high level group and a low level group based on their official distances. The cutoff official distance for the two groups was 70 m for male subjects and 50 m for female subjects.
Two high definition video camcorders were used to record the last cross step and the delivery stride of each subject at a frame rate of 60 fields/second and a shutter speed of 1/1000 seconds. Real-life three-dimensional coordinates of 21 body landmarks, the front edge of the grip, and the tail and tip of the javelin were obtained using the Direct Linear Transformation procedure. Raw coordinates were filtered through a low-pass digital filter at an estimated cutoff frequency of 7.14 Hz. The shoulder, elbow, and wrist joint angles, and the upper trunk rotation angle were reduced for analysis. The beginning times of 6 upper extremity joint and segment angular movements accelerating the javelin were identified and normalized to the double support time (time from left foot touchdown to release of the javelin).
Two-way ANOVAs with mixed design were performed for each gender to compare the sequence of the maximum linear velocity of the upper extremity joint centers and the sequences of the upper extremity joint and segment angular movements between levels of performance. A p-value of 0.05 was chosen as an indication of statistical significance.

RESULTS: The upper extremity maximum joint center linear velocities of male and female subjects were clearly in a proximal-to-distal sequence ($p < 0.001$, $p < 0.001$) (Table 1). Level did not affect this sequence for either gender ($p = 0.919$ for males, $p = 0.157$ for females). The normalized beginning times of upper extremity joint and segment angular movements of male subjects had significant sequential effects ($p < 0.001$), and were separated into 5 groups of movements (Table 2). Level did not have significant effects on the sequence of the upper extremity joint and segment angular movements of the male subjects ($p = 0.535$). The normalized beginning times of the upper extremity joint and segment angular movements of female subjects also had significant sequential effects ($p < 0.001$), and were separated into 4 groups of movements (Table 2). Level did not have significant effects on the sequence of the upper extremity joint and segment angular movements of female subjects ($p = 0.210$).

Table 1. Sequence of maximum joint center linear velocities of elite javelin throwers (% double support time) (Mean ± SD).

Joint Centers	Male		Female	
	Low level	High level	Low level	High level
Hip	0.34 ± 0.13	0.29 ±0.17	0.47 ±0.07	0.47 ±0.09
Shoulder	0.73 ± 0.08	0.73 ±0.07	0.73 ±0.07	0.70 ±0.07
Elbow	0.83 ± 0.03	0.81 ±0.04	0.82 ±0.03	0.82 ±0.03
Wrist	0.95 ± 0.04	0.93 ±0.04	0.92 ±0.02	0.94 ±0.03

Table 2. Sequence of upper extremity movements (% double support time) (Mean ± SD).

Male			Female		
Movement	Low level	High level	Movement	Low level	High level
Upper trunk forward rotation	-0.14 ±0.34	-0.15 ± 0.33	Upper trunk forward rotation	-0.07 ±0.23	-0.21 ±0.31
Shoulder horizontal adduction	0.70 ± 0.09	0.67 ± 0.10	Shoulder abduction	0.70 ± 0.15	0.72 ±0.12
Shoulder abduction	0.80 ± 0.05	0.76 ± 0.10	Shoulder horizontal adduction	0.81 ± 0.05	0.74 ±0.13
Elbow extension	0.83 ± 0.04	0.82 ± 0.04	Elbow extension	0.81 ± 0.04	0.83 ±0.04
Shoulder internal rotation	0.85 ± 0.05	0.86 ± 0.04	Shoulder internal rotation	0.83 ± 0.05	0.83 ±0.05
Wrist flexion	0.97 ± 0.03	0.97 ± 0.03	Wrist flexion	0.97 ± 0.03	0.97 ±0.03

DISCUSSION: The results demonstrated that the sequence of the maximum joint center linear velocities does not represent the sequence of the lower and upper extremity joint and segment angular movements. The results of this study showed that the maximum joint center linear velocities of javelin throwers were in a proximal-to-distal sequence while the beginning times of the upper extremity joint and segment angular motions were not in such a sequence. The results of this study also showed that the shoulder angular movements did not start at the same time, and that some shoulder joint angular movements started at the same time with elbow extension. Male and female throwers apparently used different sequences of upper extremity joint and segment angular movements. These results suggest that the upper extremity joint and segment angular movements in javelin throwing is not in a simple proximal-to-distal sequence.

Neither the sequence of the maximum joint center linear velocities nor the sequence of the beginning of the joint and segment angular movements appears to be a factor affecting the performance of javelin throwing in this study. The performances of the high and low level groups in each gender were significantly different, while the sequences of the appearances of the maximum joint linear velocities and the upper extremity joint and segment angular movements were not. Further, the variations in the sequences of the upper extremity joint and segment angular movements of the two groups in each gender were similar. These results suggest that the difference in performance between the high and low level groups in this study was due to other factors instead of the sequence of the upper extremity joint and segment angular movements.

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ASSOCIATIONS BETWEEN JAVELIN THROWING TECHNIQUE AND RELEASE SPEED

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KEYWORDS: javelin throw, technique, release speed, biomechanics.

INTRODUCTION: In javelin throwing, as with other throwing events, the release speed is the single most important factor contributing to long throws (Best et al., 1993; Hay & Yu, 1995; Hubbard et al., 2001). A greater release speed will increase the vacuum flight distance approximately in proportion to the square of the release speed (Hubbard, 1984). The vacuum flight distance is the major partial distance of official distance (Hay & Yu, 1995), and the official distance determines javelin throwing performance. Increasing release speed, therefore, will improve javelin throwing performance. The purpose of this study was to determine which technique variables are associated with greater release speeds.

METHODS: Sixty two competitive trials of 32 male and 30 female javelin throwers were recorded with two high-definition digital video camcorders placed parallel and perpendicular to the throwing direction. Twenty one body landmarks and the tip, tail, and centre of mass (COM) of the javelin were digitized for each trial. Three-dimensional (3-D) coordinates were calculated from the two camera views using the Direct Linear Transformation procedure.
The release speed of the javelin was calculated from the 3-D coordinate data of the COM of the javelin at the release. Trunk, hip, and shoulder Euler angles, the left leg-ground inclination angle, and elbow and knee angles were calculated from the 3-D coordinate data of the body landmarks. Speed and timing variables were calculated using critical instants of right foot down, left foot down, and release and the 3-D coordinate data of the body landmarks. The critical instants enabled meaningful inter-athlete comparisons to be made.
Males and females were analyzed separately to reduce confounding influences. Cross-sectional correlation and stepwise multiple regression analyses were performed to determine the relationships between the technique variables and release speed. Commonality analyses were performed to investigate interrelationships among the technique variables. A significance level of $\alpha = 0.1$ was chosen for this exploratory study.

RESULTS: For female javelin throwers greater release speeds were correlated with: shorter times in double support ($p < 0.05$); and greater runway speeds at left foot down, greater hip-shoulder separations at right foot down, and smaller left leg angles at left foot down ($p < 0.1$). A linear combination of runway speed at left foot down, right shoulder external rotation angle at right foot down, and time spent in single support accounted for 36% of release speed variability ($F = 4.79$, $p = 0.009$). Commonality analyses suggested some suppression effects and shared variance between variables (Table 1).
For male javelin throwers greater release speeds were associated with: shorter times spent in single support, greater trunk forward tilts at release, greater hip-shoulder separations at release, more right shoulder horizontal abduction at right foot down and left foot down ($p < 0.05$); and greater trunk forward tilts at left foot down, and more right shoulder external rotation at right foot down ($p < 0.1$). A linear combination of trunk forward tilt at release, left leg-ground angle at left foot down, and javelin yaw angle at release accounted for 45% of release speed variability ($F = 7.679$, $p = 0.001$). Commonality analyses suggested some suppression effects between variables (Table 2).

Table 1. Female Javelin Throwers' Release Speed Regression Equation

Variable	Mean	Coefficient	t (p)	Commonality%
Runway speed LFD	4.4 m/s	+1.602	+3.536 (0.002)	-27, -20, +13
Right shoulder ext. rotation RFD	34°	-0.018	-2.390 (0.024)	-27, -13, +13
Single support time	0.23 s	+13.113	+1.707 (0.100)	-20, -13, +13

Table 2. Male Javelin Throwers' Release Speed Regression Equation

Variable	Mean	Coefficient	t (p)	Commonality %
Trunk forward tilt REL	32°	+0.081	+3.958 (0.001)	-17, +1, 0
Left leg-ground angle LFD	46°	+0.150	+2.373 (0.025)	-17, 0, 0
Javelin yaw angle REL	-13°	-0.062	-2.347 (0.026)	1, 0, 0

DISCUSSION: Our data suggest that the technique variables associated with great release speeds are different for males and females. For female throwers the major factor for increasing release speed is to increase runway speed and maximise this speed up to the block at left foot down. Runway speed at left foot down has the strongest independent correlation with release speed and is the primary variable in the multiple regression equation. Reducing the double support phase time is a secondary factor. For a given overall throwing time any decrease in double support time will be associated with an increase in single support time. Females may also increase their release speed by applying force to the javelin for longer by increasing their throwing range of motion. This is accomplished by externally rotating their shoulder and twisting their trunk more before beginning the throwing motion, and by vaulting more during the block by planting their left foot further in front of them.

For male throwers a greater contribution from the throwing arm to generate release speed is shown. Males who are able to apply force to the javelin for longer by increasing their throwing range of motion tend to generate higher release speeds. This increased range of motion comes mainly from externally rotating and horizontally abducting their shoulder before beginning the throwing motion. Increasing trunk twist is also important. A negative correlation between release speed and yaw angle, i.e. a throw to the right side of the sector, may be reflecting this increased reliance on right arm motion to generate release speed.

Runway speed is an important factor for male throwers. In general, male throwers tend to have greater runway speeds than their female counterparts. This fact and the lack of statistically significant relationships between runway speed and release speed for males suggests there may be a threshold at which further increasing runway speed is of less importance than other factors. An increased forward trunk lean and a relatively upright left leg at the plant are not desirable, because this generates a forward pitching motion. The athlete is forced to counteract this motion driving them downwards by generating more vertical velocity, which is detrimental given the structure and line of action of human musculature. The great forward trunk leans may be indicative of an inability to control high runway speeds, which are causing forward pitching of the trunk instead of being converted to release speed.

CONCLUSION: To increase their release speeds javelin throwers should focus on increasing their runway speed, minimizing time for the throwing procedure, and effectively transferring runway momentum to release speed. Once runway speed is high, it may be more effective to increase release speed through arm motion rather than further increasing runway speed.

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ASSOCIATIONS BETWEEN JAVELIN THROWING TECHNIQUE AND AERODYNAMIC DISTANCE

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KEYWORDS: javelin throw, technique, aerodynamics, biomechanics.

INTRODUCTION: The javelin is the most aerodynamic of the four track and field throwing implements. It is twenty times more aerodynamic than the discus (Hubbard, 1984). The official distance of a throw is measured by the meet officials. The vacuum flight distance is determined by the release parameters and the range equation. The aerodynamic distance is the distance gained (where official distance is greater than vacuum flight distance) or lost (where official distance is smaller than vacuum flight distance) due to aerodynamic factors that affect the flight (Hay & Yu, 1995). The ability to gain aerodynamic distance may be the differentiating factor between athletes with near-maximal release speeds and vacuum flight distances. It may be possible for a javelin thrower to increase their aerodynamic distance independently of other performance variables, such as release speed. A javelin throwers technique determines the release parameters of the javelin, which determines the aerodynamic distance. The purpose of this study was to determine which technique variables are associated with greater aerodynamic distances.

METHODS: Sixty two competitive trials of 32 male and 30 female javelin throwers were recorded at 60 frames/second with two HDDV camcorders placed parallel and perpendicular to the throwing direction. Twenty one body landmarks and the tip, tail, and centre of mass (COM) of the javelin were digitized for each trial. Three-dimensional (3-D) coordinates were calculated from the two camera views using the Direct Linear Transformation procedure.
The release parameters of the javelin were calculated from the 3-D coordinate data of the javelin at release as described by Best et al. (1993). Release speed is the magnitude of the javelin velocity vector at release. Release angle is the direction of the javelin velocity vector at release. Inclination angle is the orientation of the javelin with respect to the horizontal. Angle of attack is the angle between the release angle and the inclination angle. Official distance was recorded for each throw and the partial distances were calculated as described by Hay & Yu (1995). Knee, hip, leg, trunk, shoulder, and elbow joint angles were calculated from the 3-D coordinate data of the body landmarks. Speed and timing variables were calculated using critical instants of right foot down, left foot down, and release and the 3-D coordinate data of the body landmarks. The critical instants enabled meaningful inter-athlete comparisons to be made.
Males and females were analyzed separately to reduce confounding influences. Cross-sectional correlation and stepwise multiple regression analyses were performed to determine the relationships between the technique variables and aerodynamic distance. Commonality analyses were performed to investigate interrelationships among the technique variables. A type I error rate of 0.1 was chosen to indicate statistical significance.

RESULTS: For female javelin throwers greater aerodynamic distances were correlated with a more extended right elbow at right foot down ($p < 0.01$). A linear combination of javelin inclination angle at release and right elbow flexion angle at right foot down accounted for 26% of aerodynamic distance variability ($F = 4.792$, $p = 0.017$). Commonality analyses suggested some shared variance between technique variables (Table 1).
For male javelin throwers greater aerodynamic distances were associated with: more shoulder adduction at left foot down, a more extended right elbow at release, and a lower javelin inclination at left foot down ($p < 0.05$). A linear combination of release angle and right

knee angle at left foot down accounted for 26% of aerodynamic distance variability ($F = 4.996$, $p = 0.014$). Commonality analyses suggested some suppression effects between technique variables (Table 2).

Table 1. Female Javelin Throwers' Aerodynamic Distance Regression Equation

Variable	Mean	Coefficient	t (p)	Commonality%
Javelin inclination REL	41°	-0.284	-2.174 (0.039)	+5
Right elbow flexion RFD	23°	-0.108	-2.084 (0.047)	+5

Table 2. Male Javelin Throwers' Aerodynamic Distance Regression Equation

Variable	Mean	Coefficient	t (p)	Commonality %
Release angle	33°	-0.547	-2.595 (0.015)	-8
Right knee flexion LFD	140°	+0.261	+2.016 (0.053)	-8

DISCUSSION: Our data suggest that the primary factor for increasing aerodynamic distance is the alignment of the javelin, specifically the inclination of the javelin above the horizontal (inclination angle). In general the javelin should be inclined relatively low to minimise the angle of attack. Female javelin throwers achieve this by maintaining extension of their elbow through the crossovers up to the throwing procedure. Male javelin throwers achieve this by maintaining extension of their elbow throughout the throwing procedure, through greater adduction of their shoulder, and from a position where their right knee is extended at the block, which suggests a longer final stride and a less upright trunk position.
A lower javelin inclination tends to decrease the angle of attack by better matching the orientation of the javelin with the orientation of the release speed, since the angle of attack is the angle between them. Release angles tend to be low, because greater release speeds can be developed at lower release angles due to the structure and preferred line of action of the human body musculature. Reducing the inclination of the javelin instead of changing the orientation of the force applied to the javelin maintains great release speeds. Matching the inclination of the javelin to the release angle minimises the angle of attack. This is important for reducing the area of the javelin that is in contact with the air, which will reduce the drag force applied to the javelin and allow for improved flight and longer aerodynamic distances. It is especially important to reduce the angle of attack where the javelin is oriented above the line of action of the force, because this will reduce the drag force applied to the underside of the javelin. Drag force applied to the underside of the javelin, particularly at the back, would cause an increased overall drag, as well as a forward pitching motion and premature drop of the point of the javelin as pressure was applied behind the centre of mass of the javelin, which acts as the pivot point.

CONCLUSION: Aerodynamic distance is primarily determined by the alignment of the javelin at release. To increase aerodynamic distance javelin throwers should lower the inclination of the javelin to match their release angles and minimise the angle of attack.

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COMING DOWN: THROWING MECHANICS OF BASEBALL CATCHERS

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Catchers are asked to make quick, highly accurate throws from a deep squat starting position. The purpose of this study was to define the throwing mechanics of catchers. Comparisons of their throwing biomechanics were made with pitching and long toss. Motion data were collected on collegiate catchers (n=8) and pitchers (n=22) making such throws in game-like situations. Catchers exhibited a significantly different stride pattern, greater elbow flexion through arm cocking, and less forward trunk tilt at ball release. The stresses on the shoulder and elbow during catchers' throws were similar to pitching and long toss, but produced significantly less ball velocity, suggesting a less efficient motion. This inefficiency is most likely compensation in order to complete the throw in less time. Coaches should be aware of this tendency when teaching catchers throwing mechanics.

KEYWORDS: compensation, efficiency, footwork.

INTRODUCTION: Catchers make more throws in a baseball game than any other player on the diamond, and many professional scouts believe throwing ability is a catcher's most important skill (Walter, 2002). While a pitcher essentially has an unlimited amount of time to prepare and make a pitch, a catcher must catch a pitched ball from a deep squat position and deliver the ball nearly 40 m in 2.0 seconds when an opposing baserunner attempts to steal second base (Walter, 2002). The pitches they receive are often thrown at a high velocity with an unknown trajectory. Rarely does the ideal condition, known as a "pitch-out", occur. In this case, the pitch is delivered at the catcher's standing chest height in the opposite batter's box. This allows the catcher to smoothly rise from his crouched position and make an unimpeded throw to second base. More commonly, a catcher must receive a low pitch thrown over the plate or a pitch in an even less desirable location, and if necessary, wait for the batter to complete his swing. As the baserunner begins advancing from first to second base, the catcher must attempt to deliver the ball as quickly and accurately as he can, using appropriate footwork and smoothly transferring the ball from glove to throwing hand. Catchers must also overcome the innate limitations of their protective gear: a facemask that obstructs their vision and a chest protector and shin guards that restrict their motion and weigh them down. Even though stolen bases are often ultimately out of their control (e.g. fast runner, slow pitching delivery, and/or poorly located pitch), many still look to the abilities of the catcher to determine the fate of the baserunner.

Most of the available literature on the throwing mechanics of catchers has been produced by coaches (Stallings, 2000; Johnson, Leggett, & McMahon, 2001). Some main coaching points are to have quick feet, align the front shoulder with the target, and have an abbreviated arm path to expedite ball release (BR) (Johnson, Leggett, & McMahon, 2001) While numerous biomechanical studies have described pitching mechanics (Dun *et. al.*, 2008; Fleisig *et. al.*, 1999; Fleisig *et. al.*, 2006; Matsuo *et. al.*, 2001; Stodden *et. al.*, 2001), only one study has attempted to quantify the throwing mechanics of catchers (Sakurai, Elliot, & Grove, 1994). Therefore, the purpose of the current study was to thoroughly describe the biomechanics of catchers' throws to second base. To put these mechanics into better context, the well-established parameters of pitching mechanics and the throwing mechanics of pitchers performing "long-toss" at a similar distance to the catchers' throws were used for comparison. Long-toss is a skilled throw used by many baseball players in practice and is commonly used in games by outfielders. It was hypothesized that because of the time demands and their initial squat stance, catchers would have significantly different mechanics than the other throwing styles. Under duress, they may sacrifice biomechanical efficiency to conserve time. Results from this study will help biomechanists and coaches better understand and teach proper throwing mechanics for catchers.

METHOD: Healthy college baseball catchers (n=8) and pitchers (n=22) were recruited for participation. Eight motion analysis cameras recording at 240 Hz (Motion Analysis Corp., Santa Rosa, CA) were initially placed in a ring around home plate. Starting from their typical crouched position, catchers received pitched balls and made ten maximum effort throws to a fielder standing at second base (approximately 40 m away). On a subsequent day, the cameras were positioned around an area along the right field foul line, and pitchers made five maximum effort “long toss” throws from flat ground at a distance of approximately 40 m to a player standing in center field. On a third day, pitchers delivered ten maximum effort pitches from an indoor mound using standard protocols (Dun *et. al.*, 2008). After signing consent forms, all players were given ample time to warm up and make as many practice throws as necessary. A total of 21 reflective markers were used to track the motions (Fortenbaugh & Fleisig, 2008), and outdoor motion capture sessions for catchers and long toss were done at night to eliminate interference from sunlight.
Relevant selected kinematic and kinetic variables were compared among the three conditions using a one-way ANOVA. Even though the same group of pitchers completed the mound pitches and long-toss throws, they were treated as though they were separate individuals because of the supposed uniqueness of the skills and to help facilitate statistical comparison. Tukey post-hoc comparisons were used to assess differences among groups. To help protect against Type I errors, α=.05.

RESULTS: All participants were approximately the same age (20.6±1.4 years) and mass (90.7±9.8 kg). However, the pitchers (187.6±6.2 cm) were significantly taller than the catchers (181.0±6.2 cm). Kinematic data for the catchers and pitchers, broken down by phase, are shown in Tables 1, 2, and 3. Kinetic data are shown in Table 4.
A number of significant differences were seen between the catchers’ throws to second base and the pitcher’s mound deliveries, most notably ball speed (36.8 $m*s^{-1}$ to 33.0 $m*s^{-1}$). At lead foot contact (FC), catchers exhibited a significantly shorter stride, more open lead foot position and closed lead foot angle, less pelvis-trunk separation and greater elbow flexion than the pitchers. The catchers maintained this greater elbow flexion throughout arm cocking and extended the lead knee more from FC to BR, but had less forward trunk tilt at BR than did the pitchers. All of these differences, except for pelvis-trunk separation at FC and forward trunk tilt at BR, were also seen comparing catchers to the long-toss throws. Catchers, like mound pitches, also had a more neutral lateral trunk tilt at FC than long-toss.

Table 1. Comparison of Throwing Kinematics at Lead Foot Contact

Variable	C	P - Mound	P – Long-Toss
**Stride length (% height)	67.1 ± 5.5	80.8 ± 4.2	81.9 ± 4.6
**Lead foot position (cm)	2.9 ± 11.7	25.5 ± 12.2	16.6 ± 14.1
**Lead foot angle (deg)	31.9 ± 5.6	14.3 ± 9.2	16.0 ± 10.8
Lead knee flexion (deg)	48.7 ± 6.2	47.7 ± 9.5	45.0 ± 9.4
Pelvis rotation (deg)	24.6 ± 16.8	36.4 ± 12.4	34.1 ± 12.0
*Pelvis-trunk separation (deg)	39.8 ± 8.9	51.5 ± 9.9	48.4 ± 11.7
**Lateral trunk tilt (deg)	2.0 ± 10.0	4.5 ± 7.0	12.3 ± 8.6
Shoulder external rotation (deg)	63.7 ± 30.9	54.5 ± 28.6	53.0 ± 28.8
**Elbow flexion (deg)	110.2 ± 15.2	79.1 ± 16.7	79.8 ± 18.1

*Significant difference among groups, p<.05.
**Significant difference among groups, p<.01.

Table 2. Comparison of Throwing Kinematics at Arm Cocking

Variable	C	P - Mound	P – Long-Toss
Pelvis rotation velocity (deg/s)	585 ± 75	569 ± 67	589 ± 60
Timing of pelvis rotation (%)	32.0 ± 19.2	25.0 ± 20.9	21.1 ± 21.7
Upper trunk rotation velocity (deg/s)	1050 ± 66	1123 ± 85	1123 ± 101
Timing of upper trunk rotation (%)	50.7 ± 15.5	50.3 ± 9.2	48.7 ± 10.8
Shoulder external rotation (deg)	175.3 ± 8.3	174.9 ± 10.8	174.3 ± 10.0
Shoulder horizontal adduction (deg)	20.0 ± 1.9	16.8 ± 6.7	18.7 ± 6.6
*Max elbow flexion (deg)	113.9 ± 12.9	99.2 ± 11.8	100.8 ± 12.1

*Significant difference among groups, p<.05.
**Significant difference among groups, p<.01.

Table 3. Comparison of Throwing Kinematics at Arm Acceleration and Ball Release

Variable	C	P - Mound	P – Long-Toss
Shoulder internal rotation velocity (deg/s)	6351 ± 761	7538 ± 1188	7288 ± 1462
Elbow extension velocity (deg/s)	2281 ± 195	2411 ± 288	2399 ± 306
*Lead knee extension (FC to BR) (deg)	19.8 ± 12.0	8.6 ± 10.5	8.2 ± 11.2
**Forward trunk tilt (deg)	23.4 ± 8.9	34.9 ± 7.6	27.5 ± 7.8
Shoulder abduction (deg)	90.7 ± 6.0	88.2 ± 7.1	88.7 ± 8.6
Elbow flexion (deg)	26.8 ± 5.3	24.1 ± 5.0	23.6 ± 5.9
**Ball velocity (m/s)	33.0 ± 1.6	36.8 ± 2.0	36.6 ± 2.1

*Significant difference among groups, p<.05.
**Significant difference among groups, p<.01.

Table 4. Comparison of Throwing Kinetics

Variable	C	P - Mound	P – Long-Toss
Shoulder proximal force (N)	1038 ± 137	1189 ± 170	1088 ± 177
Shoulder horizontal adduction torque (Nm)	94 ± 18	96 ± 18	98 ± 17
Elbow varus torque (Nm)	92 ± 17	93 ± 18	90 ± 19
Elbow flexion torque (Nm)	40 ± 9	47 ± 7	47 ± 7

*Significant difference among groups, p<.05.
**Significant difference among groups, p<.01.

DISCUSSION: It was believed that having to make an accurate long distance throw as quickly as possible would dictate that catchers have significantly different throwing mechanics than other players. Based on the results of this study, catchers clearly utilize a unique set of mechanics when making throws to second base during steal attempts. They are unable to replicate the long stride, foot placement, and pelvis-trunk separation used in long-toss and pitching. Catchers also immediately bend the elbow excessively, bringing the wrist close to the ear, and maintain this extreme position throughout the arm cocking phase. While not quite statistically significant, catchers have noticeably less rotational velocities of the upper trunk and shoulder. All of these adaptations lead catchers to have a significantly lower ball velocity despite nearly identical stress as mound pitches and long-toss on the shoulder and elbow joints. Since the catchers did not have any additional forward trunk tilt at BR, the larger amount of knee extension was attributed to them standing up out of the crouch rather than pushing the hips back to facilitate hip and trunk flexion. All comparable variables in this study were found to be similar to those reported in the Sakurai, Elliott & Grove (1994) study.

While it appears that catchers are emulating the throwing mechanics taught by coaches (Stallings, 2000; American Baseball Coaches Association, 2001), it is unclear whether a more efficient motion is feasible. It is reasonable to assume that the pitching and long-toss motions represent more biomechanically efficient alternatives, but their implementation may cost too much time to throw out runners attempting to steal second base. Further research, including measurements of total movement time and throwing accuracy, may wish to explore the possibility of different styles of throwing to determine which combines the greatest amount of biomechanical efficiency with the greatest amount of time efficiency.

CONCLUSION: Baseball catchers have a significantly different throwing motion than other positions. The most notable kinematic differences include a shorter stride, open foot position, closed foot angle, and reduced pelvis-trunk separation angle at FC; excessive elbow flexion during arm cocking; and less forward trunk tilt at BR. A clinically significant reduction in upper trunk rotation and shoulder internal rotation velocities lead to a statistically significant reduction in ball velocity, suggesting a biomechanically less efficient throwing motion than other players. It is likely that these biomechanical changes are done in the interest of minimizing the total time it takes to the deliver the ball to second base.

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ISBS 2010

Oral Session 14

Treadmill Walking & Running

THE INFLUENCE OF MANUALLY ADJUSTING THE RUNNING SPEED ON THE IMPACT ACCELERATION OF THE TIBIA DURING TREADMILL RUNNING

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The purpose of this study was to investigate the influence of pressing the speed key on the treadmill console to change the running speed on the impact acceleration of the tibia during treadmill running. Twenty-seven subjects were asked to run on the treadmill and increase the speed gradually until their preferred speed within two minutes. Then the subjects were required to adjust their running speed manually every two minutes. The peak impact accelerations of the right and left tibia were measured for 30 seconds at the 2nd, 4th, 6th, 8th, and 10th minutes. The results showed that there are greater peaks of impact acceleration of the tibia when some subjects pressed the speed key on the treadmill console to change their running speed. It is suggested that the position of the speed key of the treadmill be close to the runner.

KEYWORDS: treadmill console, preferred speed, bilateral tibial accelerations

INTRODUCTION: Because of the poor outdoor environments, indoor treadmill exercise is becoming a popular exercise to improve fitness. In previous studies, the treadmill has often been used as the experimental equipment to control conditions, such as speed and surface conditions, and to acquire the data for a continuous time (Verbitsky et. al., 1998; Derek et. al., 2002; White et. al.,2002 and Hardin et. al.,2004). Although the kinematic and kinetic characteristics have been compared between overground and treadmill running, there have been few studies to investigate the characteristics of the specific movements on the treadmill. Therefore, the purpose of this study was to investigate the influence of manual adjustments to the running speed on the impact acceleration of the tibia during treadmill running.

METHODS: Twenty-seven subjects without a history of musculoskeletal injury were recruited for this study. Fifteen subjects (S1-S15) were experienced in treadmill exercise and twelve subjects (S16-S27) were inexperienced. The subjects who had done treadmill exercise at least once a week in the past six months were defined as experienced treadmill runners. The physical characteristics of the subjects are shown in Table1. This investigation was approved by the Human Experiment and Ethics Committee of National Cheng Kung University Hospital. The subjects were informed of the experimental risks and signed an informed consent before participation.

Table1

The Physical Characteristics of the Subjects (mean ± SD)

Group	Height (m)	Mass (kg)	Age (years)
Experienced (n=15)	1.69±0.07	62.7±10.0	21.5±2.6
Inexperienced (n=12)	1.72±0.07	63.7±6.0	18.9±0.9
Total (n=27)	1.71±0.07	63.1±8.3	20.3±2.4

At first, the subjects ran on a treadmill (Figure 1, MAG7310, Tonic Fitness Technology, Inc, Tainan, Taiwan) and increased their speed gradually until they reached their preferred speed within two minutes. Then the subjects were required to manually adjust their running speed by pressing the speed key on the console (Figure 1) in order to increasing to 110% of their preferred speed, decreasing back to their preferred speed, then decreasing to 90% of their preferred speed, and finally increasing to their preferred speed every two minutes. Each testing session was finished within ten minutes. Two low-weight, three-axes accelerometers (dimensions: 33mm×28mm×19mm, weight: 17grams, range: $\pm$50g, sampling rate: 1000Hz) were attached with elastic bandages to the tuberosity of the right and left tibia for each subject. Only the peak impact accelerations in one direction along the length of the right and left tibia were measured for 30 seconds at the beginning of the 2nd (100%), 4th (100%-110%), 6th (110%-100%), 8th (100%-90%) and 10th (90%-100%) minutes. This treadmill speed has a 0.1 km/hr increment every time the speed key is pressed. The subject must press the key continually until the target speed is achieved. The time that the subject took to press the speed key was recorded by an experimenter from the time display on the console.

The data was normalized to the mean peak impact acceleration of the right tibia measured at the 2nd minute for each subject in order to remove the influences of different individual speeds and the influences of the shod conditions on the impact accelerations. Two-way repeated-measures ANOVA was used to evaluate the effects of pressing the speed key and different speed change conditions on the peak impact accelerations of the right tibia for the experienced subjects only. The SPSS version 17.0 statistical software was used ($\alpha = 0.05$). LSD (Least Significance Difference) method was used to do paired comparisons.

Figure 1: The treadmill used in this study. The circle shows the speed key position.

RESULTS: Nine subjects (six subjects in the experienced group and three subjects in the inexperienced group) in this study performed larger peak impact accelerations on the right or both tibia, but only under some speed change conditions (Table 2). Figure 2 shows the peak impact accelerations of the tibia at the 4th minute for S1 and S4. The region surrounded by the rectangle indicates the duration for which the subjects pressed the key to change running speed. It is obvious that greater peak acceleration appeared as the subjects was pressing the speed key. Statistical analysis was only used to evaluate the effects of pressing the speed key and different speed change conditions on the peak impact accelerations of the right tibia for the experienced subjects because of the greater peak accelerations occurring on the right tibia for the more experienced runners (Table 2). The results showed that significantly larger peak impact accelerations occurred in the duration of the subjects pressing the key on the console to change running speed whether under increasing or decreasing the speed conditions ($p<0.05$). However, the peak accelerations did not differ significantly under different speed conditions after pressing the speed key (Figure 3).

DISCUSSION: In this study, significantly larger peak impact accelerations appeared in the duration of the subjects manually adjusting their running speed by pressing the speed key continuously. The effect of different running speeds on impact acceleration was not significant in this study. Therefore, this indicates that pressing the speed key on the treadmill console to change the running speed perhaps interrupts the running gait and increases the possibility of larger impact accelerations on the tibia. The sudden increase in the impact acceleration perhaps causes lower extremity injuries. However, the mechanism for the influence of manually adjusting the speed movement on the suddenly increased impact acceleration is not clear. Pressing the speed key on the console was a frequent movement during the treadmill exercise. In the future, more kinematic and kinetic analysis should be involved to clarify the relationship between the movement of pressing the speed key and the gait on the treadmill.

No significant differences in peak accelerations were found for the experienced subjects running at different speeds. This means that the 10% increment or decrement of their preferred speed doesn't change the peak tibial accelerations for the experienced treadmill runners. Only three out of twelve inexperienced subjects had greater peak tibial accelerations during the time when they were pressing the speed key. Therefore, the occurrence of the sudden greater impact acceleration is not related to whether the subject is experienced in treadmill running.

Table2

The Conditions that Significant Greater Peak Acceleration of the Tibia Appeared for the Nine Subjects

Subjects	Speed change conditions (% of the preferred speed)	The tibia where the larger peak accerations appeared
S1	100%-110%	right, left
S1	110%-100%	right, left
S1	100%-90%	right, left
S4	100%-110%	right only
S5	110%-100%	right, left
S11	100%-110%	right only
S12	110%-100%	right, left
S13	100%-90%	right, left
S20	100%-110%	right only
S20	110%-100%	right, left
S21	100%-90%	right only
S26	110%-100%	right, left
S26	90%-100%	right, left

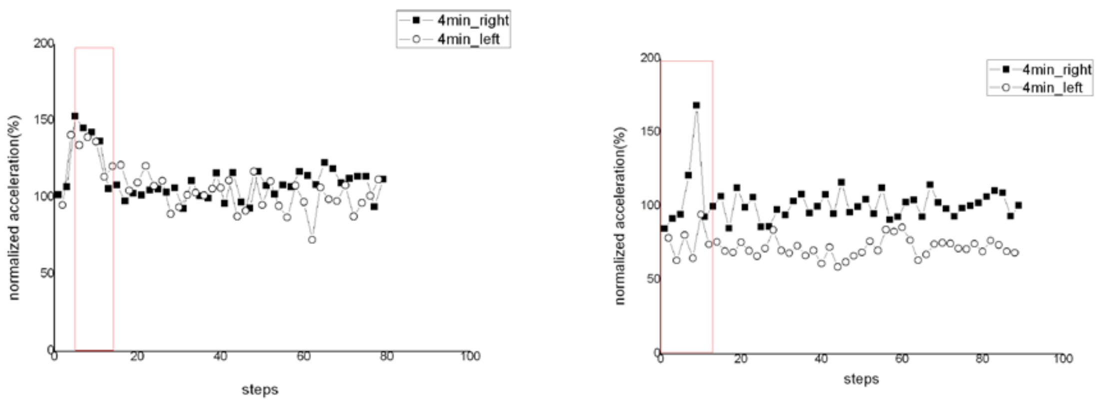

Figure 2: The peak impact acceleration of the tibia at the 4th minute for S1 (left) and S4 (right). The region surrounded by the rectangle indicates the duration for which the subjects pressed the speed key to change running speed.

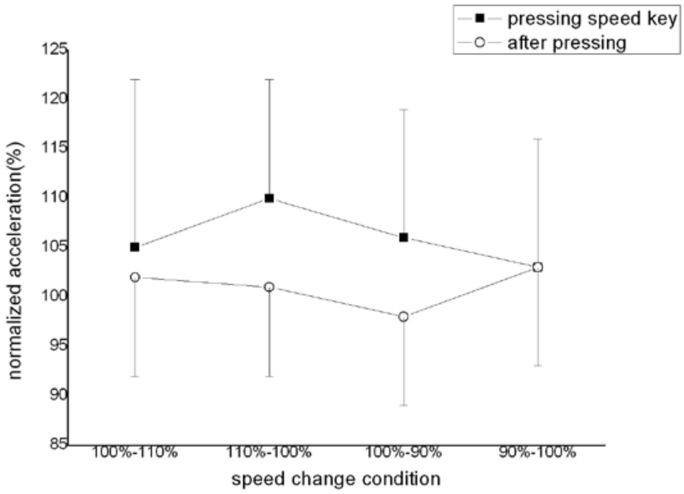

Figure 3: The mean normalized peak impact accelerations of the right tibia under eight different conditions for the experienced subjects.

CONCLUSION: Based on the results of this study, the movement of pressing the speed key seems to affect the gait of the subject during treadmill running. The movement of pressing the speed key is often performed during treadmill exercise. It is suggested that the position of the speed or incline key should be close to the runner. This mechanism should be clarified in future studies and some suggestions for key positions given.

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THE IMPACT ACCELERATION ON THE BILATERAL TIBIA DURING TREADMILL RUNNING

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The purpose of this study was to investigate the changes of the impact acceleration on the left and right tibia during treadmill running for experienced treadmill runners. The bilateral tibial accelerations of 14 subjects were measured during thirty-minutes of running at their preferred speed. Acceleration data was collected every 5 minutes for 30 seconds. There were no significant statistical differences in the peak tibial accelerations during thirty-minutes of treadmill running. The results showed that five subjects had greater right peak tibial accelerations and two subjects had greater left peak tibial accelerations. Experienced treadmill runners seemed to choose their preferred speed at that which they can run for thirty minutes and eventually there was no significant increase in the peak tibial accelerations.

KEYWORDS: bilateral tibial accelerations, experienced treadmill runners, preferred speed

INTRODUCTION: Sports activities have become increasingly popular in modern society as more and more people frequently go jogging or walking in their daily lives. A treadmill is one essential piece of exercise equipment in fitness clubs or in families. Furthermore, a treadmill has often been used as auxiliary equipment previously in studies to control the speed of the runner, studies monitoring changes in biomechanical and physiological parameters after long-term running or walking, and studies for the stability or cushion of shoes, etc. However, a treadmill was rarely considered as the major facility for investigating different model types or for cushioning effects. Such experiments may be useful in improving treadmill functions and developing new models. With the growing popularity of the treadmill, it may be even more important to perform research on treadmills in the present day.

It has been found that, with increased running time and running distance, the pressure in the sole of the foot was also increased (Verdejo and Mills, 2004). It has also been shown in this study that the long-term usage of shoes will cause structural damage to the sole, which also increases the pressure on the sole of the foot and increases the likelihood of injury. In addition, fatigue after prolonged running may weaken the cushioning effects in the lower body.

Some previous studies found that there is increased stride length, reduced frequency, and raised tibial acceleration in tired runners after long-term running (Derrick et al., 1998; Verbitsky et al., 1998). Higher tibial acceleration may also result in an increased rate of injury. However, the studies above only capture the impact accelerations of the right tibia. The purpose of this study was to investigate the changes of the impact acceleration on the bilateral tibia during treadmill running for experienced treadmill runners.

METHODS: Fifteen experienced treadmill runners were included in this study. However, the data of only fourteen subjects (age: 21.4 ± 2.74; height: 169 ± 7.2 cm; mass: 62.1 ± 9.9 kg) were acquired because the data of one subject was lost. This investigation was approved by the Human Experiment and Ethics Committee of National Cheng Kung University Hospital. The subjects were informed of the experimental risks and signed an informed consent before participation.

Two low-weight, three-axes accelerometers (dimensions: 33mm×28mm×19mm, weight: 17grams, range: +50g, sampling rate: 1000Hz) were attached with elastic bandages to the tuberosity of the bilateral tibia of each subject. Only the peak impact accelerations in one direction along the length of the right and left tibia were measured. Because thirty-minutes of exercise is good for improving human fitness, the subjects were asked to run on a treadmill (MAC-7310, Tonic Fitness Technology, Inc, Tainan, Taiwan) for thirty minutes. The subjects

had to reach their preferred speed within 2 minutes after the treadmill was started. Accelerations were measured for the 30 seconds at the beginning of the 2nd, 5th, 10th, 15th, 20th, 25th, and 30th minutes with the Acqknowledge software (BIOPAC Systems, Inc). The data has been normalized by the mean peak impact acceleration of the right tibia measured at the 2nd minute for each subject in order to remove the influences of the different individual speeds and shod conditions on the impact accelerations.
A two-way repeated-measured ANOVA was used to evaluate the differences between the peak impact accelerations of the bilateral tibia under different running times for the experienced treadmill runners with the SPSS version 17.0 statistical software ($\alpha = 0.05$).

RESULTS: The results showed that five subjects had greater right peak tibial accelerations and two subjects had greater left peak tibial accelerations, although no statistically significant differences in the averaged peak accelerations between the right and left tibia were found. Furthermore, the peak tibial acceleration did not significantly change with time (Figure 1). Figure 2 shows that subject 4 had greater right tibial accelerations, while subject 9 had greater left tibial accelerations.

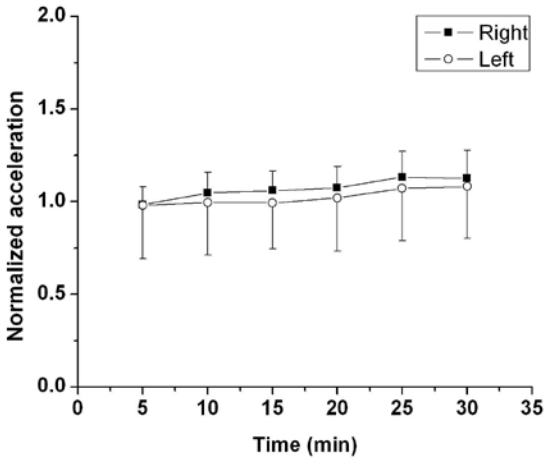

Figure 1. Mean normalized peak accelerations in 30 minutes for all subjects. There were no significant differences between the right and left tibia under different running time conditions.

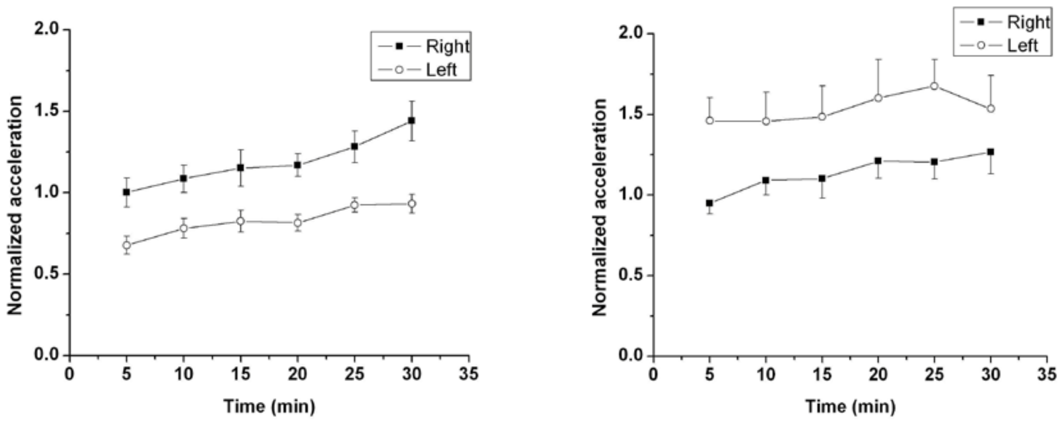

Figure 2. Mean normalized peak accelerations in 30 minutes for subject 4 and subject 9.

DISCUSSION: In the study of Verbitsky et al. (1998), 22 participants ran on a treadmill with the same shoes. The speed was increased every 30 seconds by 0.22 m/s from 1 m/s until the participant's anaerobic threshold (AT) was reached. After 30 minutes the End-tidal carbon dioxide pressure (PETCO2) test was employed to identify the subjects' fatigue level. The accelerometers were attached to the tibial tuberosity in order to collect 30 impact accelerations at intervals of 5 minutes. The peak acceleration was found to increase when fatigue occurred after 30 minutes (Verbitsky et al., 1998). The researchers postulated a significant relation between the results and the causes of running injuries. On the contrary, in this present study, the peak value did not increase with time. The experienced treadmill runners seemed to choose their preferred speed at that which they could run for thirty minutes and eventually there was no significant increase in the peak tibial accelerations.
Although there were no significant differences in the averaged peak accelerations between the right and left tibia, seven out of the fourteen subjects in this study had different impact accelerations between their right and left tibias. This means that the different running patterns of the subjects on the treadmill cause different impact accelerations to be performed on the bilateral tibia. Therefore, the bilateral tibial accelerations were suggested that they be captured under treadmill running to identify whether the maximum impact acceleration perhaps occurred in the right or left tibia.

CONCLUSION: Based on the results, there was no significant statistical difference between the peak tibial accelerations during thirty-minutes of treadmill running. The experienced treadmill runners seemed to be used to doing thirty-minutes of running to improve their fitness. Therefore, there were no increases on impact accelerations after thirty-minutes of running at their preferred speeds. The different bilateral tibial accelerations were found in half of the subjects in this study. In the research for investigating the cushioning of the treadmill's surface, a bilateral tibial acceleration measurement should be carried out to capture the real maximum impact acceleration on the right or left tibia.

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RELIABILITY OF 3D FRONTAL PLANE KNEE AB/ADDUCTION RANGE OF MOTION DURING RUNNING IN YOUNG ATHLETES

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This study quantified within-session and between-session reliability of 3D frontal plane knee ab/adduction range of motion during the stance phase of running gait calculated for 18 long term athlete development programme participants (10 males and 8 females, 11.5 ±1.4 years) during two testing sessions (spaced 10 weeks apart). Average mean differences in frontal plane knee ab/adduction between running trials (for the right or left side) within a session (week 1 or week 10) ranged from 0.2 to 7.2% (ES 0.01–0.26) which were acceptable differences. However, average mean differences between sessions for running trials (for the right or left side) ranged from 0.1 to 20% (ES 0.01–0.6). The mixed model resulted in estimates of knee ab/adduction range of motion for effects of limb side (3.6°), session (2.8°), run trial (0.2°) and subjects (4.5°). Within-session ICCs ranged from 0.80 to 0.92 and between-session ICCs ranged from 0.51 to 0.73. Based on these ICCs, within-session reliability of frontal plane knee ab/adduction is good and between-session reliability is average to good.

KEYWORDS: running, children, reliability, lower extremity, 3D kinematics.

INTRODUCTION: Screening of individuals for risk of lower limb injury and as a means to optimising performance has become common, particularly in professional sport, but also at other competitive and recreational levels (Mottram & Comerford, 2008). When assessing the lower extremity, the use of functional gait screening to evaluate movement quality is becoming common place. During assessments of gait, clinicians typically evaluate dynamic lower extremity alignment. Poor dynamic alignment has been described as a combination of excessive pelvic drop, hip adduction, internal rotation and knee valgus (Earl, Monteiro, & Snyder, 2007; Powers, 2003; Sahrmann, 2002; Willson & Davis, 2009). Poor frontal plane knee control observed during activities such as running, squatting and landing, is considered a key risk factor for the development of common injuries such as patellofemoral dysfunction. Clinically this is often observed as increased stance phase valgus angle at the knee (Powers, 2003).

Few studies have investigated the reliability of frontal plane kinematics during gait, and none have assessed children. However, it is crucial to know if kinematics are consistent enough from day to day for making clinical decisions. Reliability refers to whether a specific measurement tool produces consistent outcomes during repeated measures of the same variable (Clark, 2001). Highly sensitive sports science measurements are characterised by little variation in consecutive measures of performance (Hopkins, 2000). A change in performance due to an intervention has to be greater than the normal day-to-day training variation before coaches can conclude that the intervention has had a meaningful impact on the athlete's performance (Soper & Hume, 2004). For a performance test to be valuable it must be specific enough to measure the performance variable of interest and reliable enough to detect the relatively small differences in performances that are beneficial to elite athletes (Schabort, Hawley, Hopkins, & Blum, 1999). Utilisation of a reliable assessment tool helps ensure that variations between measurements are attributed to changes in the variable being measured (Bolgla & Keskula, 1997; Clark, 2001). Furthermore, the reliability of tests needs to be established if they are to be used in longitudinal studies evaluating injury risk or the effect of rehabilitation interventions.

The purpose of this study was to investigate within-session and between-session reliability of 3D frontal plane knee ab/adduction range of motion during the stance phase of treadmill running in healthly young athletes.

METHODS: Eighteen young athletes (10 male and 8 female, 11.5 ±1.4 years, 1.53 ±0.12 m, 44 ±7.9 kg) were recruited from an existing long-term athletic development (LTAD) programme designed to develop all-round sporting ability. All athletes were injury-free at the time of testing. Data were collected during two sessions 10 weeks apart. During each session participants underwent a treadmill-based assessment of running kinematics. A nine-camera motion analysis system (Qualysis Medical AB, Sweden) recorded lower body 3D kinematics. Twenty-one retro-reflective markers were secured to specific lower extremity anatomical locations. Two cluster marker sets (four markers attached to a plastic shell) were also attached to the thigh and shank of each leg. Children ran for five minutes at a self-selected speed (2.19 ±0.22 m/s) and kinematic data were collected in two 30-second increments at two-minute intervals. Anatomical markers were tracked using the Qualysis motion capture software and exported to Visual 3D (C-Motion Inc, USA) for calculation of relevant kinematic data. Kinematic data 'text' files were imported into Labview (National Instruments, USA) for calculation of range of motion via maximum and minimum joint angles during the stance phase of ten running strides. To summarise, each athlete completed two running trials at session 1 and session 2. Ten continuous steps for each limb were extracted from each trial for sequential analyses.
Statistical Analysis System (SAS) (SAS Institute Incorporated, USA) was used to calculate descriptive statistics including means and standard deviations (spread of results among participants) and within-session and between-session reliability of 3D frontal plane knee ab/adduction range of motion. Data were log transformed to provide measures of reliability (performance consistency) using a repeated measures analysis of variance. Reliability measures included the difference in the mean as a percentage, and Cohen's effect sizes (ES). Effect sizes are interpreted as <0.2 as trivial, <0.41 as small, 0.41-0.7 as moderate, and >0.7 as large (Hopkins, 2002). Variability measures included intra-class correlation coefficients (ICC), and typical error of measurement as a coefficient of variation percentage (Hopkins, 2000) estimated from the knee ab/adduction range of motion (discrete value). The ICC classifications of Fleiss (1999) were used to describe the magnitude of ICC values (<0.4 as poor, 0.40-0.75 as fair to good and 0.75 as excellent). A mixed modelling approach using SAS allowed quantification of both fixed effects (e.g. trial number, week of testing) and random effects (e.g. individual identity) and included variances and co-variances caused by both between- and within-subject factors (Hopkins, 2002). ICCs were calculated for a variety of steps (1 to 25).

RESULTS AND DISCUSSION: Kinematics in all three planes were measured, however, given the proposed links between poor frontal plane knee control and the development of lower extremity injuries (Powers, 2003), the focus was placed on the assessment of knee ab/adduction range of motion. Within-session descriptive and reliability statistics, including 90% confidence limits (90%CL), for knee ab/adduction range of motion for all participants are presented in Table 1. Between-session, within trial statistics for each limb are presented in Table 2. Within-session average mean differences between running trials for each limb ranged from 0.2 to 7.2%, which were acceptable differences. However, average mean differences between sessions for running trials for a given limb side ranged from 0.1 to 20%.
A standard error of measurement of 10% or less is considered small in pure test-repeats of three or more trials (Bennell, Crossley, Wrigley, & Nitschke, 1999). Our typical errors expressed as CV% were 10-13% indicating moderate variability for knee ab/adduction between subjects. Although the CV%s were moderate, the magnitude of the angles was relatively small, usually less than a few degrees. Variability in 3D kinematics may be due to errors in measurement, marker replication and movement, and variability of human locomotion. It is difficult to seperate these and therefore the variability reported in this study includes all contributions.

Table 1. Within-session statistics, including 90%CL, for frontal plane knee ab/adduction range of motion during the stance phase of running for healthly young athletes (n=18).

	Session 1		Session 2	
	Right	Left	Right	Left
Trial 1 mean ±SD (degrees)	6.8 ±2.0	6.8 ±2.9	8.1 ±2.2	6.8 ±2.4
Trial 2 mean ±SD (degrees)	7.4 ±2.5	6.8 ±2.7	8.4 ±2.6	7.2 ±2.4
ES (within-session, between-trials)	0.26	-0.01	0.10	0.15
Change in mean % (90%CL)	7.2 (0.5 to 14.4)	0.2 (-5.5 to 6.2)	2.1 (-5.1 to 9.9)	5.4 (-2.0 to 13.4)
Typical error as a CV% (90%CL)	11.8 (9.2 to 16.9)	10.5 (8.2 to 15.0)	13.5 (10.5 to 19.5)	13.4 (10.4 to 19.2)
Total error (%)	12.7	10.2	13.2	13.6
Intraclass r (90%CL)	0.87 (0.73 to 0.94)	0.92 (0.82 to 0.96)	0.80 (0.59 to 0.91)	0.90 (0.78 to 0.96)

Table 2. Between-session statistics, including 90%CL, for frontal plane knee ab/adduction range of motion during the stance phase of running for healthly young athletes (n=18).

	Right		Left	
	Trial 1	Trial 2	Trial 1	Trial 2
Session 1 mean ±SD (degrees)	6.8 ±2.0	7.4 ±2.5	6.8 ±2.9	6.8 ±2.7
Session 2 mean ±SD (degrees)	8.1 ±2.2	8.4 ±2.6	6.8 ±2.4	7.2 ±2.4
ES (for a trial, between-sessions)	0.63	0.40	0.01	0.16
Change in mean % (90%CL)	20.3 (9.4 to 32.2)	14.6 (0.7 to 30.3)	0.1 (-10.7 to 12.2)	5.4 (-7.8 to 20.4)
Typical error as a CV% (90%CL)	17.7 (13.7 to 25.6)	24.9 (19.1 to 36.5)	21.8 (16.8 to 31.8)	25.9 (19.8 to 38.1)
Total error (%)	22.8	26.7	21.1	25.5
Intraclass r (90%CL)	0.66 (0.35 to 0.84)	0.51 (0.13 to 0.76)	0.73 (0.46 to 0.87)	0.61 (0.28 to 0.81)

The mixed model resulted in knee ab/adduction estimates for effects of limb side (3.6°), session (2.8°), trial (0.2°) and subjects (4.5°). Analyses of ICCs and standard deviations (SD) expressed as degrees showed that for most variables at least 10 steps per running trial were needed. Knowledge of the variation in variables within a session and between sessions allows an estimation of the number of subjects and numbers of trials when designing experiments. For an experimental study with parallel groups (control and intervention), the number of subjects required can be determined by the equation 2*(1-ICC)*272 where the smallest worthwhile effect is 0.2 (Hopkins, 2000). The number of subjects in each group varies depending on the number of steps analysed and the subsequent ICC of the variable to be measured. For example, if three steps were analysed for a variable giving an ICC of 0.71 then the equation 2*(1-0.71)*272 results in 158 subjects in each group. If ten steps were analysed for a variable giving an ICC of 0.78 then the equation 2*(1-0.78)*272 results in 119 subjects in each group. If ten steps were analysed for a variable giving an ICC of 0.96 then the equation 2*(1-0.96)*272 results in 22 subjects in each group.

A change and/or reduction in frontal plane knee motion when running is potentially important for designing injury prevention interventions. It is therefore essential that we have a good appreciation of how reliably this can be measured. These results demonstrate that knee ab/adduction can be reliably measured within acceptable limits both within-sessions and between-sessions. However, it should be noted that to achieve this level of reliability at least 10 steps should be analysed.

CONCLUSION: Within-session and between-session reliability of knee ab/adduction range of motion during the stance phase of running in a young athlete population demonstrated average to good reliability. Knee ab/adduction range of motion could be a useful clinical screening tool.

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CHANGES IN STEP LENGTH AND WIDTH DURING TREADMILL RUNNING

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The purpose of this study was to measure a runner's step length and width during thirty-minutes of treadmill running. The step kinematics for sixteen subjects, ten experienced and six inexperienced treadmill runners, were acquired and analyzed. Four LED markers were positioned on the front tips of the toe boxes and heel counters on both the right and left shoes at the same level. The positions of the four markers were captured by a Visualeyez system for ten seconds at the beginning of the 5 th and 30 th minutes. The results showed that the subjects seemed to increase their step length backward in their toe-off position after thirty-minutes of treadmill running. The maximum range of movement, defined as the product of the maximum step length and the maximum step width, was 66% and 30% of their height for the subjects during thirty-minutes of running.

KEYWORDS: treadmill deck, motion capture, experienced treadmill runners

INTRODUCTION: Because of poor outdoor environments, indoor treadmill exercise is becoming a popular exercise to improve the fitness. In previous studies, the treadmill has often been used as the experimental equipment to control conditions, such as speed and surface, and to acquire data for a continuous time (Verbitsky et. al., 1998; Derek et. al., 2002; White et. al., 2002 and Hardin et. al.,2004). Although the ground reaction forces were compared for experienced and inexperienced treadmill runners, the differences in the kinematical characteristics, such as changes in step length and step width, during long time treadmill running have not been determined. The purpose of this study was to measure the runner's step length and width during long time running. The maximum range of movements on the treadmill was calculated in order to give reference to the adequate area of the treadmill deck.

METHODS: Twenty-seven subjects without a history of musculoskeletal injury were recruited in this study. However, only the kinematical results of sixteen subjects, ten experienced and six inexperienced treadmill runners, were acquired and analyzed. The subjects who had done treadmill exercise at least once a week in the past six months were defined to be experienced treadmill runners. This investigation was approved by the Human Experiment and Ethics Committee of National Cheng Kung University Hospital.
At first, the subjects ran on a treadmill (MAC-7310, Tonic Fitness Technology, Inc, Tainan, Taiwan) and increased their speed gradually until their preferred speed within two minutes. Because thirty-minutes of exercise is good to improve human fitness, the subjects were asked to run on a treadmill for thirty minutes. Four LED markers were positioned on the front tips of the toe boxes and heel counters on both the right and left shoes at the same level (Figure 1). A Visualeyez system that consists of two trackers was used to capture the three dimensional positions of the markers automatically. In order to compute the coordinates of a marker, the positions of the two trackers should be adjusted before capturing the data to make sure that all three sensing eyes of the tracker can see the marker simultaneously. The positions of the four markers were captured for the ten seconds at the beginning of the 5 th and 30 th minutes. In this study, the step length was determined by the distance between the maximum forward and backward positions of the four markers and the step width was determined by the distance between the maximum right and left positions of the four markers. Then, the movement range was defined as the product of the step length and the step width. The kinematic data was collected by the VZSoft and smoothed using a 4th-order Butterworth low-pass filter at a cut frequency of 6Hz. All the data was filtered and analyzed by the Matlab 7.0 program. In order to compare the differences between the experienced and inexperienced groups, the two variables were normalized to subject's height.

Figure 1. There are two markers on the front tips of toe boxes and heel counters of each shoe. The positions of the four markers are at the same level (arrows).

RESULTS: The results of the experienced subjects (Table 1) and inexperienced subjects (Table 2) indicated that thirteen subjects (eight experienced and five inexperienced) increased their step lengths after thirty-minutes of running. By comparing the step widths at the 5th minute, nine subjects (five experienced and four inexperienced) enlarged their step width at the 30th minute. Although there was no statistical analysis in this study, the mean normalized step length increased for both the experienced (46% increase to 52% of height) and inexperienced groups (48% increase to 51% of height). The increased mean step width (17% increase to 20% of height) was only found for the inexperienced subjects. Excluding Subject 15, however, the differences in the step width are not significant for the other inexperienced subjects.

According to the results for all subjects, the maximum range of movement for the experienced group altered from 59%*23% to 66%*26% of their height, and the inexperienced group changed from 58%*22% to 60%*30% of their height. Therefore, the maximum range of movement was 66%*30% of their height for the all subjects during the thirty-minutes of running on the treadmill.

Table 1. The Step Length and Step Width for the Experienced Subjects

	5 th minute				30 th minute			
	SL(cm)	SW(cm)	SL/H(%)	SW/H(%)	SL(cm)	SW(cm)	SL/H(%)	SW/H(%)
S1	78.0	26.0	46	15	96.6	25.1	57	15
S2	74.0	29.3	42	17	94.1	26.1	53	15
S3	61.9	27.5	34	15	76.6	30.4	43	17
S4	98.1	33.7	59	20	109.2	32.3	66	20
S5	92.0	29.4	52	16	98.8	30.4	55	17
S6	82.2	37.8	51	23	81.6	42.6	50	26
S7	77.9	27.4	49	17	66.2	27.9	41	17
S8	56.4	31.4	34	19	86.5	23.9	52	14
S9	74.3	38.3	45	23	91.8	41.0	55	25
S10	72.0	29.5	45	18	74.9	24.6	47	15
Mean (s.d.)	76.7 (12.4)	31.0 (4.3)	46 (8)	19 (3)	87.6 (13.0)	30.4 (6.6)	52 (7)	18 (4)

Note: SL:Step Length; SW: Step Width; SL/H: the ratio of Step Length to Height; SW/H: the ratio of Step Width to Height

Table 2. The Step Length and Step Width for the Inexperienced Subjects

	5 th minute				30 th minute			
	SL(cm)	SW(cm)	SL/H(%)	SW/H(%)	SL(cm)	SW(cm)	SL/H(%)	SW/H(%)
S11	75.1	33.2	44	19	80.3	33.4	47	19
S12	103.8	26.6	58	15	108.7	24.7	60	14
S13	86.9	31.5	52	19	88.0	32.7	52	19
S14	73.1	38.0	43	22	89.8	35.6	53	21
S15	84.3	25.1	49	15	90.2	51.5	52	30
S16	68.9	22.5	44	14	61.8	23.7	40	15
Mean	82.0	29.5	48	17	86.5	33.6	51	20
(s.d.)	(12.7)	(5.8)	(6)	(3)	(15.3)	(10.0)	(7)	(6)

Note: SL:Step Length; SW: Step Width; SL/H: the ratio of Step Length to Height; SW/H: the ratio of Step Width to Height

DISCUSSION: The purpose of this study was to investigate the changes in the step length and step width after thirty-minutes of treadmill running. Based on the results, most of the subjects' step lengths were more likely to lengthen (80% of experienced and 83% of inexperienced). The subjects seemed to take off the treadmill deck at a position farther back to increase step length, since their landing position did not change a lot (Figure 2). The increased step length would be accompanied by more ankle plantar flexion.

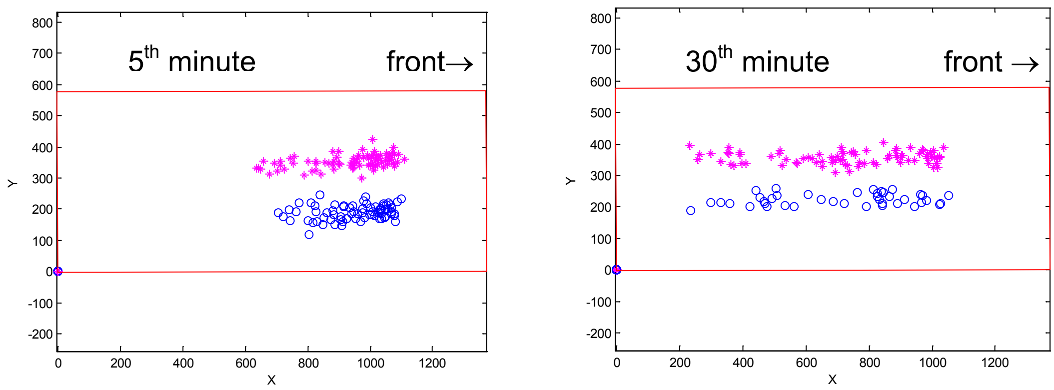

Figure 2. The changes of step length and step width after thirty-minutes of running for a subject. The red frame is the region of the treadmill deck. The right side of the figure is the frontal direction of the treadmill.

A greater percentage of inexperienced subjects (67%) had an extended step width after thirty-minutes of running than experienced subjects (50%). However, excluding subject 15, the step widths had less increase. This result showed that subject 15 had unstable steps after 30 minutes of treadmill running. However, from the evidence it was not obvious to infer that the inexperienced treadmill runners would extend their step width after thirty-minutes of running because of the fewer inexperienced subjects that were included in this study.

CONCLUSION: In this study, the range of the movement on the treadmill increased after thirty-minutes of running for both the experienced and inexperienced subjects. The subjects seemed to increase their step length backward in their toe-off position. An inexperienced subject was found to have a significantly increased step width after thirty-minutes of running. More investigations need to be done to identify whether the inexperienced treadmill runners have unstable steps after a long period of running. Based on the results of this study, the maximum range of movement was 66%*30% of their height for the subjects after thirty-minutes of running. The data will be a good reference in order to produce a treadmill that has an adequate area of its deck for the safety of leisurely users.

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IDENTIFICATION OF EMG FREQUENCY PATTERNS IN RUNNING BY WAVELET ANALYSIS AND SUPPORT VECTOR MACHINES

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The purpose of this study was to identify EMG pattern of running at different speed and incline based on a trial-to-trial analysis. Eight subjects performed treadmill running at five different conditions (4, 5 and 6 m/s, 5m/s at 5° incline, 5m/s at 2° decline). EMG data of eight leg muscles were recorded and transformed by a wavelet analysis (van Tscharner, 2000). Ten subsequent steps of each subject and condition were classified by support vector machines. Between 93 and 100% of all EMG patterns were assigned correctly to the individual. According to the different running conditions recognition rates ranged between 78 and 88%. Hence, support vector machines can be considered as powerful nonlinear tool for the classification of dynamic EMG patterns.

KEYWORDS: pattern recognition, support vector machines, electromyography, running.

INTRODUCTION: The electromyographic activity of a single muscle is considered as a complex stochastic signal that results from the superposition of the electrical activity of several motor units and therefore shows a high trial-to-trial variability. In dynamic movements such as running, the coordination of muscles with similar function (e.g. extension of the knee) can enhance this variability since the same movement outcome might be produced by different activities of single muscles. The most common approaches for the analyses of the surface electromyogram (EMG) aim on the extraction of the essential contents of the signal by averaging over time and trials (De Luca, 1997; Hermens et al 1999). In running, such techniques have been applied to analyze the EMG of the leg muscles at different speed (e.g. Kyröläinen et al. 2005, Gazendam & Hof 2007) and incline (e.g. Swanson & Caldwell 2000). However, there are two major disadvantages of these techniques. Calculating the signal mean over time (e.g. by root mean square) allows quantifying the overall intensity but does not provide any process-related information about the changes of EMG intensity during performance. If the EMG is averaged over trials, signal variations from trial to trial are primarily considered as noise and therefore neglected. It is implicitly assumed that differences between groups of trials (e.g. according to varying running conditions) must reflect in the average activity of a single muscle. Variations of the EMG that result from compensatory activities of different muscles cannot be obtained and hence the interplay between muscles cannot be analyzed.
In this paper a different approach was chosen that takes benefit of the variability of the EMG signal and considers the interplay in muscular activity between different muscle groups. The main objective was to identify EMG patterns for running at different speed and incline.

METHOD: Eight track and field athletes (age 18.6 years ±2.4; 5 male/3 female) participated in this study. All subjects were free from recent lower extremity injury or pain and trained regularly for at least 2 years. After a five minute warm up the subjects were asked to run five times 200 m on a treadmill at different conditions (4, 5 and 6 m/s, 5m/s at 5° incline, 5m/s at 2° decline). Recovery periods between the trials were chosen by the subjects individually and lasted normally approximately one minute. EMG of eight muscles of the right limb were recorded at 2400 Hz using bipolar surface electrodes (AMBU 720 00-S) and single differential amplifiers (BIOVISION). The following muscles were considered: M. gastrocnemius medialis, M. lateralis, M. soleus, M. tibialis anterior, M. biceps femoris, M. rectus femoris, M. vastus medialis and M. vastus lateralis. The raw signals of each EMG channel were amplified by a factor of 2000. A band-pass filter with a bandwidth from 10 to 700 Hz was applied. An accelerometer was fixed on the subjects' right shoe and was used to

determine the time of heel strike by a rapid change in acceleration. All data were collected synchronously and stored on a PDA that the subjects carried on their back (fig. 1). For each condition, the EMG of ten consecutive double steps were cut off and analyzed step by step.
For the analysis of dynamic contractions, the non-stationarity of the EMG signals must be considered as this might cause errors in the time as well as in the frequency domain. Therefore, the EMG data were preprocessed by a wavelet analysis (van Tscharner, 2000). The wavelet-transformation is a suitable method to analyse non-stationarity bio-signals simultaneously in time- and frequency-domain. The wavelet transformation of an EMG Signal is performed as convolution of the signal with the wavelet. A filter bank of 11 non-linearly scaled wavelets was used that has been especially developed for EMG application (van Tscharner, 2000, 2002).
All-in-all 378 movement patterns were analyzed. Every single movement pattern is represented by a *n x D* matrix, where *n* is the number of acquired data vectors during the stride length and *D* is the dimension of the data vectors. Each data vector consists of 88 features, as every EMG of the 8 muscles was transformed into 11 wavelets. In order to cope with the huge amount of data, dimension reduction was necessary. Dimension reduction tries to eliminate redundancy from the data by so called feature extraction. The dimension *D* is mapped to a lower dimension *d* while trying to retain the geometry of the data as much as possible. Several linear and nonlinear methods exist for this purpose. Two possibilities can be applied to reduce the matrix in which a single movement pattern is stored. The first possibility is to reduce the dimension of the features. This will lead to a matrix with *n* acquired data vectors that transport some encapsulated information on the whole movement. In the other case the dimension reduction is conducted over the time. This technique is for example used in gait analysis with kinematic position data (Troje, 2002). For example, if a Principal Component Analysis is applied to the data, the movement is transferred into a low-dimensional space spanned by the first (few) so-called eigenpostures of the walker. Similar to this approach multidimensional scaling (MDS; Cox & Cox, 1994; Kruskal, 1964) was used as a nonlinear reduction technique. MDS tries to retain the pair wise distances between the data points as much as possible during the mapping of the data to a lower dimension. The quality of the dimension reduction is expressed as a so-called stress function that is a measure of the error between the pair wise distances in the low-dimensional and high-dimensional data space. The main goal of the mapping process is hereby the minimization of the stress function (Cox & Cox, 1994). In several tests an intrinsic dimension of 4 was estimated as optimal using Matlab and a proper toolbox (Van der Maaten, 2007). Hence, the data matrices were reduced to 4 x 88 matrices. A second (so called two-fold) reduction was omitted in first instance.
After dimension reduction support vector machines (SVM; Vapnik, 1995; Chang & Lin, 2001) were used for the classification of the movement patterns. SVMs are supervised machine learning methods used for classification and regression, dealing successfully with small datasets and finding global minima (Bennett & Campbell, 2000). After compulsory amplitude normalization of the data, SVMs were trained with the bigger part of the data linked with the associated class memberships (i.e. person; running speed), and tested with the remaining data in order to calculate rates of how well those patterns were linked with the correct classes, that were excluded from the training process. This was conducted using cross validation (Jain, Duin & Mao, 2000), a standard technique to ensure more precise recognition rates and to avoid overtraining.

RESULTS: Table 1 shows the recognition rates for EMG patterns according to the individual subject. Overall, the recognition rates ranged between 92,9% and 100%. Best results were found for level running at different speeds, where all EMG pattern were assigned correctly to each subject. Similar recognition rates were found if only the two different incline conditions were considered.

Table 1. Recognition of Individual EMG Patterns

Condition	Recognition rate
Level running at 4, 5 and 6m/s (238 trials)	100%
Running at 5m/s [+5°/±0°/-2°] (220 trials)	97.7%
Slope running [+5°/-2°] (140 trials)	99.3%
All trials (378 trials)	92.9%

Recognition rates observed for the running speed and incline conditions are listed in Table 2. About 88% of all EMG patterns were classified correctly if all trials at a running speed of 5 m/s were analyzed. This includes level running as well as runs at an incline of 5° and a decline of -2°. This sample shows an even better recognition rate as for the incline conditions only (82,1%). Lowest rates were achieved for level running. Here, about 78% of all trials were assigned to the correct running speed.

Table 2. Recognition of Speed and Incline

Condition	Recognition rate
Level running at 4, 5 and 6m/s (238 trials)	78.6%
Running at 5m/s [+5°/±0°/-2°] (220 trials)	88.2%
Slope running [+5°/-2°] (140 trials)	82.1%

DISCUSSION: The single EMG patterns embody highly individual characteristics that remain stable for each subject more or less independent from speed and incline. The individual recognition rates are far beyond chance level and reach values that have been reported in previous studies for less variable kinematic data (e.g. Jaitner et. al. 2001). Moreover, specific patterns for different running speeds and inclines can be identified with high probability. This is even more remarkable since the incline differs only slightly from level running. It is therefore assumed that the muscular activity of the leg muscles during running adapts very sensitively to environmental changes.

From a methodological view, two specific aspects of the muscular activity can be addressed within this analysis, that at the first sight seem contrarily: a high individuality of muscular activity pattern that seems widely independent from changes in speed and incline and on the other hand specific pattern that remain stable for certain conditions. This highlights some critical aspects of traditional approaches in EMG analysis described in the introduction. Overall, support vector machines can be considered as powerful nonlinear tool for the classification of dynamic EMG patterns.

If the EMG patterns for a specific running condition (e.g. running at a speed of 5m/s) remain stable within the subject but differ substantially between subjects this has considerable impact on the interpretation of EMG data. Practical implications that result from the comparison of different subjects might be misleading and should be drawn with particular care. Hence, approaches that focus on the analysis of multiple trials of the same subjects might be more reliable.

An emphasis of this study was on the complex interplay between different muscles in running. The results indicate that compensatory muscular activities could be a key factor for the overall stability of the EMG patterns. The analysis of the interaction between various leg muscles therefore might provide a more detailed insight in the mechanism of running coordination. However, further research is needed to allow a better understanding of the intermuscular coordination in complex movement patterns.

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Oral Session 15

Walking & Running

CHANGE IN FOOTSTRIKE POSITION IS RELATED TO ALTERATIONS IN RUNNING ECONOMY IN TRIATHLETES

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KEYWORDS: running, kinematics, triathlon

INTRODUCTION: Biomechanical factors are likely related to the impairment in running economy frequently observed in triathletes when running after cycling (Millet et al., 2000). Cycling has been shown to interfere with muscle recruitment during subsequent running in some highly-trained triathletes (Chapman et al., 2008), but the implications of this on run performance are unknown. Links between muscle recruitment and running economy have been established during isolated running (Paavolainen et al., 1999), which compel the proposition that any change in muscle recruitment following cycling might be associated with running economy. Stride frequency, stride length and hip and knee angles have been reported to be unchanged after cycling (Quigley & Richards, 1996; Hue et al., 1997), however, muscle recruitment and limb movement have not been simultaneously measured in previous studies that have investigated the relationship between biomechanical factors and running economy after cycling. The purpose of the current investigation was to evaluate changes in neuromuscular control (muscle and movement control) during running after a 45 min high-intensity cycle and their relationships to alterations in running economy.

METHOD: Seventeen moderately-trained triathletes participated. Running economy and neuromuscular control were determined by measuring submaximal VO_2, lower limb electromyography (EMG) and sagital plane kinematics for 4 min at 12 $km.hr^{-1}$ during a control run (no prior cycling) and a run after 45 min of cycling (transition run).
A Pearson's correlation was performed to examine the univariate relationship between changes in EMG, kinematics, and VO_2 for control-transition-run comparisons. Significant univariate variables were retained and entered into a stepwise logistic regression model to determine the most accurate set of variables for prediction of an alteration in VO_2 following cycling. A backward elimination regression analysis was also performed to confirm the contribution of the variables to the model. A significance level of 0.05 was necessary to enter the variable into the model and 0.10 required for the removal of the variable to minimise the likelihood of excluding potentially useful variables.

RESULTS: Eight triathletes demonstrated a clinically meaningful (i.e. > 2.4% change as previously described by Saunders et al., 2004a) increase or decrease in VO_2 during the transition run. Correlation analysis of whole group data revealed that four kinematic variables of knee angle at foot contact, ankle angle at foot contact, total excursion of motion at the knee and minimum excursion at the knee were significantly associated with the change in VO_2 following cycling. Subsequent stepwise linear regression revealed that the change in ankle angle at foot contact was most explanatory, with backward elimination revealing the change in ankle angle at foot contact alone explaining 67.1% of the variance in VO_2 compared to 77.5% for all four variables. The association between ankle angle at foot contact and VO_2 was positive; with an increase in ankle-dorsiflexion (relative to the control run) associated with an increase in VO_2 during the transition run.

DISCUSSION: We found the angle of the ankle at foot contact was most closely related to the change in running economy, explaining 67% of the variance in VO_2. A shift to a more dorsi-flexed ankle and extended knee at foot strike increases vertical ground reaction forces

(GRF) (Gerritsen et al., 1995) and vertical GRF are major determinants of metabolic cost during running (Saunders et al., 2004b). An increase in ankle dorsi-flexion angle at foot contact, or a tendency to heel-strike, reduces conversion of translational energy into rotational energy as most of the energy is lost in collision with the ground (Lieberman et al.,, 2010). In contrast, it has been speculated that landing in a more plantar-flexed position may enhance performance through exploitation of elastic energy storage and conversion (Liebermanet al., 2010).

CONCLUSION: Changes in kinematics at the knee and ankle were most correlated to alterations in VO_2 after cycling, with the angle of the ankle at foot contact being predominant. Therefore ankle position at ground contact may be important for triathlete peformance and training interventions aimed at restoring running kinematics after cycling may benefit some triathletes performance.

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ANALYSIS OF THE BACKPACK LOADING EFECTS ON THE HUMAN GAIT

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KEYWORDS: gait, load, backpack, ground reaction force.

INTRODUCTION: Gait is a simple activity of daily life and one of the main abilities of the human being. Often during leisure, labour and sports activities, loads are carried over (e.g. backpack) during gait. These circumstantial loads can generate instability and increase biomechanical stress over the human tissues and systems, especially on the locomotor, balance and postural regulation systems. According to Wearing (2006), subjects that carry a transitory or intermittent load will be able to find relatively efficient solutions to compensate its effects. These are dependent upon the walking distance and of the load characteristics - size, weight and location relatively to the body (Hsiang, 2002).Thus, these solutions should become a concerning factor (Koh, 2009) and a topic of scientific research, particularly in what concerns the inventory of its biomechanical effects and the possible strategies to be developed in order to minimize its effects.
The aim of the present study was to analyze the effects of an occasional dorso-lombar load during the gait through the use of a backpack.

METHOD: Data was collected from forty healthy subjects (twenty males: mean stature 1.75±0.07m and mass 72.01±6.75kg; twenty females: mean stature 1.63±0.06m and mass 59.45±5.71kg), students of Sport Sciences with body mass index (BMI) less than 25, aged between 18 and 45 years and without any dysfunction that affect the independent gait. The subjects were informed of the purpose of study and all signed written informed consent.
Gait characterization was accomplished through ground reaction force (GRF) analysis. To collect the GRF data, a BERTEC force plate (model: 4060-15) was used. A devoted amplifier system (BERTEC AM 6300) and a 16 bits analogical-digital conversion unit (BIOPAC) were also used. The sampling rate was established at 1000 Hz.
Each subject was assessed initially in a normal condition (without load) and then loaded (backpack condition). Data were collected regarding three valid rehearsals of each test on the force plate (right foot). Subjects carried on the backpack, fixed at the dorso-lombar region, a static load that allow the subject + backpack to reach the “total BMI” of 30. Each subject walked three times in each condition, at a self-selected velocity, along the experimental walkway (600 cm × 92 cm × 15 cm) in which the force plate was engraved.
The results for the three components of the GRF (vertical, anterior-posterior and medial-lateral) were expressed as percentages of the total weight, with and without load. The statistical analysis was conducted with SPSS 16.0 software. Data on independent variables studied were statistically analyzed by measures of central tendency (mean) and dispersion (standard deviation), and compared by paired student t-test (with vs. without load), the significance level adopted was α=0.05.

RESULTS: The main results of the study are presented in Table 1. Results include chronometric (temporal) and the dynamometric (GRF) variables. Statistical significant differences ($p < 0.05$) are marked (*). From these we highlight an increase in the stance phase duration and a reduction in the relative magnitude of the first and second peaks of the vertical component in the load condition.

DISCUSSION: During loading, an increase in the two peak values of the vertical GRF component was observed, together with an increase of the stance phase duration. A higher value of the horizontal braking force (anterior-posterior GRF component) was also noted. On the contrary, a reduced maximal value of the latero-medial component was registered. These results showed that even when the weight of the backpack is included in the calculations of GRF as a percentage of total weight (body + pack), the mean values obtained with and without load traduce a relevant disturbance of the dynamometric profile of the gait pattern. These results conflict with previous results from Tilbury-Davis and Hooper (1999), which evaluated the biomechanical effects of load (20 and 40 kg) in military subjects. Nevertheless, those were trained subjects in this particular task.

Table 1. Mean and standard deviation (Std.) values for the studied chronometric and dynamometric variables obtained for unloaded and loaded situations in both genders

Variables	Normal		Loaded		Confidence Interval			
	Mean	Std.	Mean	Std.	Lower	Upper	t	Sig.
Duration stance phase (s)	0.78*	0.06	0.81*	0.07	-0.055	-0.014	-3.36	0.002
First peak - Vertical Component								
(N/BW)	1.03*	0.04	0.99*	0.06	0.021	0.057	4.35	0.000
Time (% stance phase)	25.96	3.09	26.68	3.24	-1.674	0.200	-1.59	0.120
Minimum value between Vertical peaks								
(N/BW)	0.82	0.05	0.82	0.06	-0.016	0.019	0.18	0.859
Time (% stance phase)	46.10	5.95	46.02	4.32	-1.740	1.896	0.09	0.931
Second peak - Vertical Component								
(N/BW)	1.10*	0.05	1.07*	0.06	0.018	0.052	4.20	0.000
Time (% stance phase)	74.64*	2.46	72.64*	3.40	1.139	2.865	4.69	0.000
Braking Force - anteroposterior component								
(N/BW)	-0.14*	0.03	-0.15*	0.03	0.003	0.018	2.93	0.006
Time (% stance phase)	18.12	2.66	17.99	1.87	-0.655	0.956	0.38	0.708
Propulsion Force - anteroposterior component								
(N/BW)	0.19	0.03	0.18	0.03	-0.001	0.014	1.79	0.081
Time (% stance phase)	83.09	1.91	82.99	1.90	-0.473	0.664	0.34	0.736
Peak - mediolateral component								
(N/BW)	0.10*	0.02	0.09*	0.01	0.002	0.012	2.77	0.009
Time (% stance phase)	46.91	21.91	46.32	18.51	-4.898	6.081	0.22	0.829

* Statistical significance $p < 0.05$

Pierrynowski, Norman and Winter (1981) suggested that gait adjustments occurred when loads of 34kg are carried by subjects whose weight was approximately 72kg (47% body weight). The maximal load used in our research for all the studied subjects was lower than the critical absolute value of 30kg proposed by the referred authors, but sometimes higher in relative terms considering the subject's body weight (57%), which suggest the need of a revision of the boundary proposed by the referred authors. Authors also reported an attenuation of the loading and unloading rates when carrying the higher load of 40kg, suggesting a protection of the biomechanical system. This seems to be in agreement with the reduction of the relative values for the first and second peaks of the vertical GRF component for the load condition obtained in our study, combined with the increased stance duration.

CONCLUSION: The present study showed an adaptation of the subjects to the load condition, with: (i) an increase of the stance phase duration; (ii) a significant reduction of the GRF vertical component peaks, and (iii) an increase of the horizontal braking force.

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CUSHIONING OF THE RUNNING SHOES AFTER LONG-TERM USE

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The purpose of this study was to investigate the cushioning properties of the running shoes after long running distance. Each of five subjects wore a new Nike air-shox shoe at the beginning and then at least ran thirty minutes on the same treadmill once or twice a week. The results of material test showed that impact force peaks significantly increased as the running distance increased. However, in the subject test, the tibial peak accelerations decreased as the running distance increased. It seemed to indicate that the subjects accommodate themselves to the material characteristics of the testing shoe by reducing the impact energies as heel strike. Based on the results, the cushioning abilities of the running shoes were attenuated after 300 km running distance. In the future, the change of the cushioning abilities of the running shoes should be monitored after more running distances.

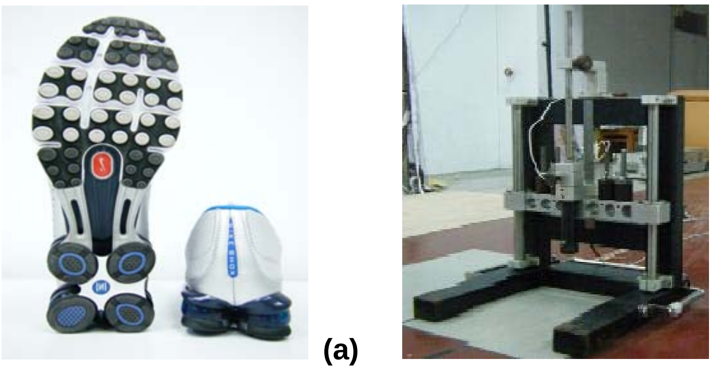
KEYWORDS: ground reaction force, subjects test, material test, tibial acceleration.

INTRODUCTION: Running and jogging are the most popular recreational activities in the world. However, the consecutive impact shocks due to foot strike may cause the chronic musculoskeletal injuries of lower limb. The cushioning properties of running shoes may play an important role to avoid running injuries. In previous studies, two methods were used to evaluate the cushioning properties of shoes: material test and subject test. The results of the two tests were conflicted in many studies (Kaelin et. al., 1985；Foti & Hamill, 1993；McNair and Marshall, 1994). The authors of these studies suggested that the material test is not valid to evaluate the situation of actual subject running. But in Chiu's study (2000), varying impact weight and impact height of the striker was used to test the cushioning of the shoe. The results showed that the curves of vertical GRF during the initial impact phase in subject test were similar to the results of material test. Chiu recommended that varying the impact energy in the material test to correspond with the impact energy of human running could evaluate the cushioning properties of running shoes validly.
Past studies concerned about the cushioning ability of running shoes after long distance running have showed that structural damage occurred in the foam of the midsole (Verdejo & Mills, 2004) and the shock absorbing ability would reduce (Schwanitz & Odenwald, 2008) after long distance running. Kinematics has been found to change to adapt shoe degradation during long distance running, especially in ankle joint movement (Kong et al., 2008). Furthermore, Lafortune and Hennig (1992) indicated that tibial axial acceleration was more sensitive at distinguishing the cushioning of different footwear than ground reaction force measurement. It has been shown that the impact acceleration and knee flexion angle would increase with fatigue (Verbitsky et al., 1998; Mizrahiet al., 2000). Therefore, in order to understand the interaction between the subject and shoe, the present study used both material test and subject test to investigate the change of cushioning abilities of the running shoes after a long running distance.

METHOD: Five recreational runners (Table 1) were recruited in this study. Before data collection each subject signed an informed consent, which was approved by the Human Experiment and Ethics Committee of National Cheng Kung University Hospital. Each subject wore a commercial running shoe (Nike, air shox 318684-142, US size 6.5-10.5) as shown in Figure 1a and finished a thirty minutes running session on the same treadmill (SportsArt 631) in the fitness gym nearby the laboratory once or twice a week. After each running session, each subject had to record the running time, distance and speed and return the testing shoe to the laboratory. All the testing shoes were preserved under humidity-controlled environment in the laboratory.

Table 1. The Characteristics of the Five Subjects

Subject	Shoe size (US)	Age (yrs)	Speed (km/hr)	Height (cm)	Weight (kg)	Gender
S1	6.5	21	10	161	52	Female
S2	10.5	24	10	178	70	Male
S3	9	29	8.5	171	61	Male
S4	9.5	25	10	180	70	Male
S5	8	24	10	165	62	Male



(a) (b)

Figure 1. (a) The shoe tested in this study, and (b) the portable impact tester.

The portable impact testing equipment (see Figure 1b) was used to impact the running shoes on a force plate (AMTI BP400600). The impact testing with potential energy ranging from 0.61 to 6.08 J (equally distributed) was performed on the shoe after every 100 km running distance. The ground reaction forces were measured at a sample rate of 1000 Hz, and the signals were filtered using a 100Hz low-pass filter. Mean ground reaction force peaks were calculated from five impacts under each impact energy condition after omitting two extreme values.

The impact accelerations of the right tibial as heel strike were acquired during 30 minutes treadmill running after every 100 km running distance for each subject. A low-weight, three-axes accelerometer (dimensions: 33mm×28mm×19mm, weight: 17 grams, range: ±50g, sampling rate: 1000Hz) was attached to the tibial tuberosity of the right leg by elastic tape. The axial direction of the accelerometer was along the tibial longitudinal axis. Each subject was asked to run on a treadmill (MAC-7310, Tonic Fitness Technology, Inc, Taiwan) in the laboratory and increase the speed gradually until the same speed of the running session in two minutes.The acceleration data were acquired for 10 seconds at the 2^{nd}, 5^{th}, 10^{th}, 15^{th}, 20^{th}, 25^{th} and 30^{th} minute.

Two- way repeated measures ANOVA with the statistical software (SPSS, v17.0) was used to identify the effects of impact energies and running distance on the GRF peaks for material test and identify the effects of running time and running distance on the peak impact accelerations of the right tibial for subject test (α=0.05). For each subject, the peak impact accelerations were normalized by the mean peak acceleration of the 2^{nd} minute at 100 km running distance.

RESULTS: The mean time of finishing the running distances of 100, 200, 300km for the five subjects were presented in Table 2. The averaged 8 ~ 10 km running distance per week was similar to that of the general runners. The results of material test showed that the vertical GRF peak increased significantly as the impact energy increased ($p < 0.05$). The GRF peak also significantly increased as running distance increased ($p < 0.05$) (Table3). For subject test, there were no significant differences of the peak impact accelerations under different run distances and running time (Table 4).

Table 2. The Time (Mean ± S.D.) of Finishing the Running Distances of 100, 200 and 300km for the Five Subjects

variable	100 km (n=5)	200 km (n=5)	300 km (n=5)
time (weeks)	11.8±2.5	21.6±3.2	38.4±5.3

Table 3. The Vertical GRF Peaks (Mean ± S.D.) in Impact Testing (Unit：N)

Impact Energy (joule)	0 km (n=5)	100 km (n=5)	200 km (n=5)	300 km (n=5)
0.61	206.6±42.1	208.7±44.4	227.6±14.5	235.1±24.4
1.22	339.3±26.8	359.4±34.0	368.1±28.6	382.8±22.6
1.82	452.9±30.3	472.9±43.4	486.3±36.9	507.4±32.4
2.43	553.4±36.3	577.1±49.0	590.3±35.5	601.9±34.4
3.04	643.0±39.2	664.3±47.0	674.7±26.0	685.3±32.9
3.65	723.3±31.2	746.4±44.4	758.2±34.0	756.1±31.2
4.26	798.8±45.0	808.4±47.3	826.2±33.2	830.7±33.7
4.87	856.2±48.7	876.4±43.2	895.5±40.5	898.4±34.4
5.47	921.0±48.3	921.6±25.6	954.1±36.0	953.4±35.4
6.08	985.2±46.9	987.4±26.0	1016.2±31.6	1023.6±39.1

Table 4. The Normalized Peak Impact Accelerations (Mean ± S.D.) of the Right Tibial in Running Test

Running time (min)	100 km (n=5)	200 km (n=5)	300 km (n=5)
2	1.00±0.0	0.98±0.20	0.98±0.29
5	1.05±0.17	0.97±0.24	0.94±0.29
10	1.15±0.31	1.00±0.29	0.95±0.28
15	1.08±0.24	1.12±0.35	1.03±0.28
20	1.12±0.28	1.02±0.29	0.99±0.26
25	1.00±0.13	0.98±0.32	1.09±0.31
30	1.11±0.27	1.07±0.40	1.08±0.24

DISCUSSION: After 300 km running distance, the results of material test showed that impact force peaks increased significantly under different impact energy conditions. This indicated that the cushioning abilities of the running shoes were attenuated after long-term use. In previous studies concerned with the cushioning ability of the running shoes after long distance run, the running distance ranged from 500 to 750 km (Verdejo & Mills, 2004; Kong et al., 2008). The total running distance in this study was smaller than those of the past studies, the

significant deterioration of cushioning ability of the running shoe should be detected after more running distance.
Although there were no statistical differences, the results of subject test showed that the peak accelerations decreased as the running distance increased. It seemed to indicate that the subjects accommodate themselves to the material characteristics of the testing shoe soles by reducing the impact energies as heel strike.

CONCLUSION: Based on the results of the material test, the cushioning ability of the running shoes significantly decreased after 300 km running distance. However, the peak impact accelerations of the tibia had a slightly reduction after 300 km running distance. The subjects seemed to adjust their landing strategies to reduce the impact energy responding to the attenuated cushioning abilities of the testing shoes. In the future, the change of the cushioning abilities of the running shoes should be monitored after more running distances.

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EFFECTS OF BACKWARD WALKING AS A MODALITY FOR LOW BACK PAIN REDUCTION IN ATHLETES

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The therapeutic effectiveness of backward walking for treatment of low back pain (LBP) was examined among athletes experiencing LBP and healthy non-athletes. All participants were pre-tested walking backward, performed 10-15 mins of backward walking three days/week for three weeks and were post-tested. Low back sagittal and coronal plane range of motion, shock attenuation (SA), stride length (SL), stride rate (SR), velocity and LBP were evaluated (α=0.05). All variables were significantly different between groups, excluding SA. Velocity, SL and SR were significantly different pre vs post. Owing to the clinical nature of this study, single-subject analyses were performed and identified unique individual responses to the intervention. Results suggest that backward walking may assist some athletes presenting with LBP.

KEYWORDS: athletic rehabilitation, retro walking, single-subject, spine, treatment modalities

INTRODUCTION: The number of collegiate athletes who experience LBP is thought to range between 1% to greater than 30% (Spencer & Jackson, 1983; Watkins & Dillin, 1999; Bono, 2004). Bountiful radiographic evidence exists documenting the fact that vertebral disc degeneration is higher in athletes than in non-athletes (Bono, 2004) yet no known cause-effect relationship has been advanced. Sports which consist of repetitive back hyperextension motion such as diving, gymnastics and wrestling have been reported to be associated with higher rates of spondylolysis (Wier & Smith, 1989; Hodges & Richardson, 1999; Bono, 2004), yet the occurrence of these vertebral stress fractures has not been documented to be higher in athletes versus non-athletes.

For many athletes, LBP can be highly debilitating, leading to reduced or total elimination of training and their focus shifts to rehabilitation. Varying degrees of intensity of LBP coupled with individual pain tolerance levels can allow some athletes to continue to train while managing LBP with non-surgical or non-physician directed therapy. Effective modalities to reduce the limited training and downtime associated with LBP are continuously being sought by athletic trainers. Backward walking (BW) and running have been promoted anecdotally and have shown potential kinematic benefits (Bates et al., 1986) but have not received thorough scientific scrutiny with respect to LBP. Masumoto et al. (2007, 2009) have shown that BW in a water environment increased core muscle activity and metabolic cost versus walking forward in the same environment. Whitley and Dufek (2009) documented increased flexibility of the hamstrings following a BW intervention. We sought to expand on these findings and questioned the effectiveness of BW as a modality for relief of LBP in athletes.

The primary purpose of this study was to investigate the effectiveness of a BW exercise program in alleviating LBP in athletes. A secondary purpose was to identify which aspects of BW performance may be beneficial (if any) in alleviating LBP. We hypothesized that the use of BW as an intervention would reduce athletes' self-reported LBP measures and modify impact attenuation charactistics and low back/pelvic movement displayed during BW.

METHOD: Five NCAA Division I athletes currently experiencing LBP (21.2±5.1yrs, 172.8±7.3cm, 68.5±7.7kg) and five active, healthy collegiates free from LBP (21.6±1.5yrs, 168.1±7.0cm, 63.0± 0.6kg) volunteered to participate in the study. Inclusion criteria for the LBP group was stipulated as having experienced LBP in the past 8 months and currently electing to allow the pain to resolve without physician involvement for the present time and the duration of the study. All granted written consent in accordance with policies established for the Protection of Human Subjects at the affiliated university.

Data were obtained from all study volunteers both pre- and post intervention. Pre and post-testing consisted of first subjectively reporting a LBP pain value (LBP group only). All volunteers practiced walking backward on a treadmill (Precor, Model C966) prior to data collection. When each participant reported that they felt comfortable walking backward without use of external support (grasping treadmill rails), BW velocity was established and encouraged to be as fast as comfortably possible. Participants were then instrumented with two lightweight uniaxial accelerometers (PCB Piezotronics Inc., Model 352C68), one secured to the distal anterior surface of the right tibia and the other to the midpoint of the forehead. A biaxial electrogoniometer (Biometrics, Model SG150) was secured externally to the low back, spanning T12-S2. Participants then walked backward on the treadmill for nine minutes with data obtained synchronously (1000 Hz) using Bioware data acquisition software (Kistler, version 3.21) during the sixth minute of the walk (Melcher et al., 2008). Following the pre-test, participants completed three weeks of supervised BW on a treadmill for 15 mins/day, three days/week. Following completion of the intervention, all participants were post-tested following the same procedures as the pre-test.

Ten strides per participant-condition were extracted from the continuous data sets using the tibial accelerometer time-history profiles to define each stride. Accelerometer and electrogoniometer data were filtered with a 4^{th} order low pass Butterworth filter (20 Hz) using a custom laboratory program (MatLab version 6.1). Dependent variables (DVs) included walking velocity (Vel), subjective pain measure (P; LBP group only), shock attenuation (SA: [1-(peak head acceleration/peak leg acceleration)]*100), stride length (SL), stride rate (SR), sagittal plane range of motion (sROM) and coronal plane range of motion (cROM) of the low back. The mean values of ten footfalls per participant-condition-test session were utilized for two (group) x two (time) mixed model analysis of variance procedures for each DV ($\alpha=0.05$). P scores for the LBP group were evaluated using a correlated t-test. Owing to the clinical nature of the investigation, we also sought to explore results of the LBP participants on a single-subject statistical basis and did so using the Model Statistic technique (Bates et al., 2003) for SA, sROM and cROM to explore potential systemic kinetic or low back kinematic changes.

RESULTS: Group descriptive data are summarized in Table 1. There were no significant group x time interactions for any of the DVs. A significant decrease ($p=0.004$) in P for the LBP group was observed following BW. Significant group differences were observed for Vel ($p<0.0001$), sROM ($p=0.0067$), cROM ($p=0.0487$), SL ($p=0.0002$) and SR ($p=0.0012$) while significant time effects (i.e., pre vs. post) were observed for Vel ($p=0.0004$), SL ($p=0.0110$) and SR ($p=0.0213$). SA was not different nor were there any significant changes in low back kinematics across time. The Model Statistic single-subject analysis procedure identified a significant reduction in SA (7.8-24.0%) following the intervention for four of the five participants ($p<0.05$). There was a significant increase in sROM (4-6 deg) and cROM (3-12 deg) observed for three of the five participants following the intervention ($p<0.05$).

DISCUSSION: LBP is a complex clinical presentation for the athletic trainer and is thought to be best managed by categorizing or matching treatments to particular symptomology (Heck and Sparano, 2000). There was no attempt in the current work to categorize specific LBP etiology nor to control or suspend other forms of treatment modalities for participants. Despite these limitations, results identified significant gait-related changes for both groups following BW intervention. Both groups increased velocity, stride parameters, and low back ROM following three weeks of BW exercise. It appears that the presence of LBP did not interfere with the ability of participants to adapt to BW. Both groups achieved greater walking velocity with a greater percent increase in SL vs SR. In order to explore the possible relationship between SL and sROM, we examined the ratio of percent change in sROM:SL. This ratio was 0.89 for the healthy (control) group and 1.26 for the LBP group, possibly suggesting that increased SL was achieved with a greater change in sROM for the LBP

group vs. the healthy group. Importantly, all LBP participants reduced self-reported P and over half significantly increased low back ROM, suggesting, as has been previously reported (Whitley and Dufek, 2009) that BW may improve low back and hamstrings flexibility.
During BW, hip extension and knee flexion is greater than in forward walking (Yang, et al., 2005). Greater hip extension and a concomitant extension of the lumbar spine increasingly load the facet joints opening up the disc space, causing a reduction in compressive loads to the intervertebral discs (Heck and Sparano, 2000). This unloading of the discs may be a mechanistic outcome of BW via increased hip extension as evidenced by the decreased P scores reported by the LBP group. As well, increased loading of the facet joints may explain the increased low back ROM observed for both groups.

Table 1. Descriptive results (mean ± standard deviation) by group-time and pre-post % change.

DV	Healthy (n=5)			LBP (n=5)		
(units)	Pre	Post	% change	Pre	Post	% change
Pain	na	na	na	3.2	2.0*	-37.5
(integer)				1.3	1.0	
Velocity	0.54†	0.87 *†	61.1	1.14†	1.38*†	21.1
(m/s)	0.04	0.21		0.13	0.11	
SL	0.79†	1.08 *†	36.7	1.22†	1.36*†	11.5
(m)	0.13	0.14		0.08	0.15	
SR	0.69†	0.81*†	17.4	0.94†	1.02*†	8.5
(hz)	0.07	0.10		0.05	0.13	
SA	67.3	69.8	3.7	64.5	52.0	-19.4
(%)	26.0	17.3		11.6	19.9	
sROM	4.6†	6.1†	32.6	11.0†	12.6†	14.5
(deg)	2.3	1.8		1.6	6.0	
cROM	14.3†	18.8†	31.5	9.7†	13.1†	35.0
(deg)	4.9	5.9		2.3	7.4	

Note: Mean followed by standard deviation; See text for abbreviations; % change=[(post-pre)/pre) * 100]; na=not applicable; *=significant difference ($p<0.05$) between conditions (Pre, Post); † = significant difference ($p<0.05$) between groups (Healthy, LBP).

Single-subject analysis results provided additional insight into the effects of the intervention. SA is a measure that captures a sense of how the body attenuates shock generated at impact due to foot contact with the ground (Mercer et al., 2002). In the current study, the increase in BW velocity resulted in a decrease in SA for 4 of the 5 participants. Peak leg acceleration (LgPk) did increase with increased BW velocity; however, it was coupled with a concomitant increase in peak head acceleration (HdPk). HdPk has been shown to remain relatively unchanged at 1.0-2.0 g's during forward running (Mercer et al., 2002, Dufek et al, 2008). In the current study, both LgPk and HdPk increased with an increase in BW speed during the post-test for most participants, while HdPk values remained well below 1.0 g (range=0.4-0.7 g's). Interestingly, one individual with an increase in SA for the post-test (6.4%) exhibited the smallest increase (0.14 m/s) in BW velocity of all study participants. This might suggest a different BW strategy for this subject in order to accommodate the LBP. Single-subject kinematic outcomes also provided insight into possible adaptation strategies. Three participants significantly increased sROM (average=4.8 deg) and cROM (average=6.6 deg) with one participant significantly reducing sROM (7.5 deg) and cROM (3.2 deg). BW appeared to significantly increase low back motion and reduce LBP for three individuals while one individual appeared to adopt a unique BW strategy.
Limitations of the study do not allow one to state definitively that BW only led to the observed outcomes of reduction in self-reported LBP, increased walking velocity, and increased low back ROM for most participants. No control was imposed upon individuals relative to supplementary forms of treatment, with the exception of physician intervention.

Symptomology was not screened and categorized (Heck and Sparano, 2000). Time itself may have contributed to the reduction in LBP. Despite these limitations and in light of the debilitating effects that LBP can produce for an athlete, we suggest further study into functional changes that may be elicited as a result of BW.

CONCLUSION: Study results present trending evidence in support of BW relative to pain reduction and increased low back ROM for athletes with LBP. Single-subject evaluation provided insight into possible mechanistic changes elicited by the BW for specific individuals with LBP, including an increase in SL accompanied by increased sROM. Clearly, additional research into the effects of BW is warranted for athletes presenting with unresolved LBP.

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Oral Session 16

Training

ELECTROMYOGRAPHICAL ANALYSIS OF LOWER EXTREMITY MUSCLE ACTIVATION DURING VARIATIONS OF THE LOADED STEP UP EXERCISE

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This study evaluated the biceps femoris, gluteus maximus, gluteus medius, rectus femoris, semitendonosus, vastus lateralis, and vastus medialis activation during four variations of the step up exercise. The exercises included the step up, crossover step up, diagonal step up, and lateral step up. Fifteen women who regularly engaged in lower body resistance training performed the four exercises with 6RM loads on a 45.72cm plyometric box. Data were collected with a telemetered EMG system, and RMS values were calculated for EMG data for eccentric and concentric phases. Results of a repeated measures ANOVA ($p \leq 0.05$) revealed a variety of differences in muscle activation between the exercises.

KEYWORDS: gluteus medius, program design, ACL injury, women

INTRODUCTION: Quantification of muscle activation of lower body resistance training exercises allows practitioners to make informed decisions regarding which exercises are optimal for performance enhancement and rehabilitation. The hamstring muscle group is important in reducing ACL injury risk and training reduces hamstring inhibition and quadriceps to hamstrings ratio (Ebben, et al., 2009). While there is a growing body of literature on hamstring activation during resistance exercise and hamstring to quadriceps ratios, few have examined the eccentric and concentric phases (Wright, et al.,1999) or the role of the gluteus medius in closed chain resistance exercise (Ayotte, et al., 2007; Ekstrom, et al., 2007; Worrell, et al., 1993). Though data has indicated reduced firing of gluteus maximus during single leg activities (Zazulak, et. al., 2005), little data exists to describe the role of the gluteus medius. It is suggested that gluteus medius training improves both strength and timing of gluteus medius firing, which may reduce dynamic knee valgus during sport and exercise, reducing risk of ACL injury (Myer, et al., 2004).

Instead, research has commonly focused on the thigh musculature during variations of the step up exercise. The primary focus of previous studies has been the rehabilitation of the knee, with experimental procedures based on commonly utilized rehabilitation protocols such as step heights of 8 inches or lower (Ayotte, et al, 2007; Beutler, et al., 2002; Ekstrom, et al, 2007; Kerr, et al., 2007), and only body weight resistance (Ayotte, et al, 2007; Beutler, et al, 2002; Bolgla, et al, 2008; Brask, et al, 1984; Childs, et al, 2004; Cook, et al, 1992; Ekstrom, et al, 2007; Kerr, et al, 2007), thereby applying rehabilitative loads and conditions to non-rehabilitation populations. Those studies that did utilize additional resistance when assessing the step up loaded subjects arbitrarily with body weight plus an additional 25 percent of the subject's body weight (Selseth, et al, 2000; Worrell, et al, 1993; Worrell, et al, 1998) out of concern for the limited capacity of rehabilitation patients rather than using either RM testing or predictive regression tools (Ebben, et al., 2008). However, the existing literature has shown the benefits of using loaded single-leg exercises to improve functional and sport performance in athletes (McCurdy & Conner, 2003), since progressive overload is necessary (Fleck & Kraemer, 1997).

The purpose of this study is to examine muscle activation during 4 variations of the loaded step up exercise using prescribed 6RM loads to determine hip and knee muscle activation.

METHODS: Fifteen women (mean ± SD; age 21.0 ± 1.41 yr; body mass 63.56 ± 6.89 kg, height 159.84 ± 28.99 cm) volunteer university students who regularly engaged in lower body resistance training served as subjects. The study was approved by the institution's internal review board. All subjects performed a habituation and testing session. Prior to each session, the subject warmed up and performed dynamic stretching. During the habituation session, all subjects were familiarized with the test procedures, including performing maximum voluntary isometric contractions (MVIC)s recorded in order to normalize the electromyographic (EMG) data. During this period, rectangular shaped, bipolar EMG surface electrodes with 1 x 10 mm 99.9% Ag conductors and an inter-electrode distance of 10 mm were placed on biceps femoris (BF), gluteus maximus (GMx), gluteus medius (GMe), rectus femoris (RF), semitendonosus (ST), vastus lateralis (VL), and vastus medialis (VM). Data were recorded using a four channel, fixed shielded cabled, DelSys Bagnoli-4 EMG system (DelSys Inc., Boston, MA, USA.) and an Elgon goniometer (DelSys Inc., Boston, MA, USA.). MVICs for the BF and ST groups were measured at 60 degrees of knee flexion using the seated leg curl (Hammer Strength, Schiller Park, IL, USA), at 60 degrees of knee flexion for the VL, VM, and RF on the leg extension machine (Magnum Fitness Systems, South Milwaukee, WI, USA), with subject lying prone at approximately 70 degrees hip flexion on a decline bench for the GMx (Magnum Fitness Systems, South Milwaukee, WI, USA), and GMe was tested with subject's leg abducted to approximately 25 degrees against a padded, immovable mass. Subjects also received instruction in and performed the four exercises including the step up (SU), crossover step up (CR), diagonal step up (DI), and lateral step up (LA). Subjects were then tested in order to determine their six-repetition maximum (6RM) for each step up variation. Six RM loads were chosen since this study sought to test muscle strength as opposed to muscle endurance. Approximately 72 hours after the habituation session, subjects returned for the testing session. During the testing session, subjects performed the same dynamic warm up session as in the habituation session, followed by 5 minutes of rest. Subjects then performed 2 repetitions of each of the step up test exercises in a randomized order with 6RM load, with 5 minutes of rest between each exercise. These exercises were selected for evaluation since they all are characterized by hip and knee extension, and DI, LA, and CR are additionally characterized by hip ab- and adduction in a dynamic, single-leg fashion, which is thought to elicit greater GMe activation (Kraus, et al., 2009).

The statistical analyses were undertaken with SPSS 17.0. A two way mixed ANOVA with repeated measures for step up exercise type was used to evaluate the main effects for step up variation and the interaction between step up variation and eccentric/concentric phase, for RMS EMG of the SU, CR, DI, and LA. Data were expressed as a percentage of MVIC for each muscle group. Bonferroni adjusted pairwise comparisons were used to identify the specific differences in muscle activation for each exercise. Assumptions for linearity of statistics were tested and met. An *a priori* alpha level of $P \leq 0.05$ was used with post hoc power and effect size represented by d and η_p^2, respectively.

RESULTS: The analysis of EMG data revealed significant main effects ($p \leq 0.001$) for BF, GMx, GMe, RF, ST, and VL, but not for VM ($p=0.833$). Analysis revealed no significant interactions between exercise type and phase ($p \leq 0.05$) for the BF, GMx, RF, ST, VL, VM. A significant interaction ($p \leq 0.001$) was found for exercise type and phase for GMe.

Table 1. RMS EMG data for 7 muscles during eccentric and concentric phases of 4 step up variations (N=14)

		SU	CR	DI	LA
BF	Eccentric phase	0.032 ± 0.015[a*]	0.031 ± 0.015[c*]	0.027 ± 0.019[c*]	0.019 ± 0.009[b*,c*,d*]
	Concentric phase	0.092 ± 0.052	0.080 ± 0.047	0.090 ± 0.055	0.070 ± 0.038
	a= significantly different from LA		*c= significantly different from DI*		**= p ≤ 0.05*
	b= significantly different from SU		*d= significantly different from CR*		
GMx	Eccentric phase	0.040 ± 0.034	0.105 ± 0.297	0.036 ± 0.022	0.032 ± 0.018
	Concentric phase	0.098 ± 0.143	0.053 ± 0.024	0.061 ± 0.029	0.064 ± 0.047
GMe	Eccentric phase	0.042 ± 0.020[a*,b*]	0.039 ± 0.021[a*,b*]	0.040 ± 0.022[b*,c*,d*]	0.038 ± 0.023[a*,c*,d*]
	Concentric phase	0.070 ± 0.028	0.077 ± 0.035	0.065 ± 0.022	0.054 ± 0.022
	a= significantly different from DI		*c= significantly different from SU*		**= p ≤ 0.05*
	b= significantly different from LA		*d= significantly different from CR*		
RF	Eccentric phase	0.054 ± 0.019[a*,b*]	0.053 ± 0.018[a*,b*]	0.062 ± 0.019[c*,d*]	0.060 ± 0.022[c*,d*]
	Concentric phase	0.084 ± 0.024	0.087 ± 0.020	0.092 ± 0.030	0.093 ± 0.030
	a= significantly different from DI		*c= significantly different from SU*		**= p ≤ 0.05*
	b= significantly different from LA		*d= significantly different from CR*		
ST	Eccentric phase	0.046 ± 0.024[a*,b*]	0.036 ± 0.012[c*,d*]	0.039 ± 0.015[a*,b*]	0.028 ± 0.011[c*,d*]
	Concentric phase	0.093 ± 0.036	0.071 ± 0.033	0.089 ± 0.039	0.069 ± 0.028
	a= significantly different from CR		*c= significantly different from SU*		**= p ≤ 0.05*
	b= significantly different from LA		*d= significantly different from DI*		
VL	Eccentric phase	0.116 ± 0.066	0.104 ± 0.067	0.110 ± 0.061[a*]	0.099 ± 0.053[b*]
	Concentric phase	0.183 ± 0.099	0.186 ± 0.104	0.191 ± 0.110	0.178 ± 0.083
	a= significantly different from LA		*b= significantly different from DI*		**= p ≤ 0.05*
VM	Eccentric phase	0.084 ± 0.044	0.085 ± 0.044	0.088 ± 0.047	0.079 ± 0.037
	Concentric phase	0.144 ± 0.077	0.145 ± 0.073	0.150 ± 0.085	0.144 ± 0.081

SU= Step up	LA= Lateral Step up	GMe= Gluteus medius	VL= Vastus Lateralis
CR= Crossover Step up	BF= Biceps femoris	RF= Rectus femoris	VM= Vastus medialis
DI= Diagonal Step up	GMx= Gluteus maximus	ST= Semitendonosus	

DISCUSSION: This is the first known study to use systematically tested RM loads to analyze EMG activity of the GMe musculature along with other hip and thigh musculature during variations of the loaded step up exercise. Significant differences were found between exercises as well as between concentric and eccentric phases for the GMe, contrary to findings of Ayotte, et al., (2007) who found no significant differences in GMe activation between front step up and lateral step up exercises in unloaded subjects. Specifically, the crossover step up was found to elicit the greatest concentric activation of the GMe, while the step up elicited the greatest eccentric activation, which we conclude was due to the starting position of CR, which placed the lead leg of the subject into femoral adduction. As a result, GMe showed greater activation during the concentric phase of the CR, as the position likely forced the muscle to fire in an attempt to abduct the femur. This finding suggests the CR should be included in resistance training programs for court and field sport athletes in an attempt to reduce incidence of dynamic knee valgus, a common injury position due to unplanned changes of direction and cutting maneuvers (Hewitt, et al., 2010). In this study, the GMx showed no significant differences in activation regardless of exercise, suggesting similar strengthening effects as determined by Ayotte, et al. (2007) during various single leg exercises. In the current study the RF, interestingly, showed greatest activation during the LA and DI exercises, both of which were performed with relatively lighter loads when compared to the SU and CR. Significant differences were found for the hamstring musculature (BF, ST) during concentric and eccentric phases of the step up variations, with more activation occurring during SU and DI up variations. It is suspected that the requirement of more sagittal plane movement of the limb coupled with the advantageous line of pull of the hamstrings in that position increase activation. Activation levels for the BF and ST were relatively low when compared to VL and VM musculature for the selected exercises, consistent with existing literature (Ayotte, et al., 2007; Brask, et al., 1984; Cook, et al., 1992; Isear, et al., 1997). The VL and VM showed no significant differences between concentric and eccentric phases, contrary to findings of Selseth and colleagues (2000), who found significant differences in activation between concentric and eccentric phases for the LA exercise.

CONCLUSION: There are several practical applications that can guide the use of variations of the step up exercise for maximal muscle activation. For maximal GMe activation, the CR should be used, while the SU and DI should be used for maximal hamstring activation. To best activate the rectus femoris, the LA and DI should be utilized. Ultimately, it appears that a varied resistance program employing all variations of the step up exercise would be the most effective approach in maximally activating the hip and thigh musculature.

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SIX WEEK CONSISTENCY OF SENSORIMOTOR TEST METHODS

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The purpose of this study was to compare sensorimotor testing methods. Therefore 15 healthy and sporty subjects undertook five different sensorimotor tests and repeated the same tests six weeks later without executing any specific sensorimotor training. The main outcome was that movement unspecific and simple tests like the Counter Movement Jump, the maximum isometric force and rate of force development on a blocked leg-press or a single-leg-stance with closed eyes have a better retest-reliability than more specific movements like a balance test on a balance board or a complex movement like a single-legged jump landing. Tests with a specific movement show a learning effect and at complex movements there was almost no correlation, as slight changes in the motion sequence can lead to big differences in the measured scores.

KEY WORDS: repeatability, control group, sensorimotor testing

INTRODUCTION: Many different testing methods for both the sensorimotor system (see Riemann et al. (Riemann and Lephart, 2002) for an overview) and postural balance (see Huxham et al. (Huxham et al., 2001)) exist. Although there is a huge choice of different testing methods; some of the methods have only been tested by the developer of the test or the manufacturer of the testing device. Some methods aim to reflect the improvement of the neuromuscular system whereas others aim to test static or dynamic balance. However, there are no gold-standards yet to test improvements in sensorimotor and balance skills caused by sensorimotor training (SMT). Recently it has been shown, that the rate of force development (RFD) can be improved by SMT (Gruber and Gollhofer, 2004). The RFD is defined as the steepness of the force-time-curve and is an important parameter to express the explosive force of the neuromuscular system. The isometric RFD of the leg extensors can be measured with a force plate fixed to a blocked leg-press. More recently it has been shown that also the maximum isometric force, which can also be measured on a blocked leg-press, can be improved by SMT (Bruhn et al., 2006). Hence, these findings indicate that the RFD method might be considered as an outcome measure for training studies of the neuromuscular system, provided it has acceptable reliability. Another method to assess improvements of the neuromuscular system is the measurement of the jump height. Therefore different jump trials can be used, like the counter movement jump (CMJ), Squat Jump or the Drop Jump at which the maximal jump height is measured. All three jump forms have been shown to be improved by SMT (Taube et al., 2007). A more functional testing method for dynamic stability that is often used to measure ankle or knee instabilities is the measurement of time-to-stabilization (TTS) or the medio-lateral displacement at a single-legged jump-landing on a stable surface (Gribble and Robinson, 2009; Ross et al., 2005; Wikstrom et al., 2005). However, jump height and test execution differ from study to study. The measurement of the displacement of the Centre of Pressure (CoP) in a single-leg-stance without previous performance of dynamic postural tasks is applied to quantify the postural sway in static standing position. The MFT S3-Check is a testing device for dynamic standing stability on an unstable support surface, which shows good reliability, objectivity and validity according to a study of the manufacturer (Raschner et al., 2008). The MFT-Board is a board that can be tilted up to 12° to the left or the right side or from forward to backward, depending on the standing position on the board.
The aim of this study was to determine which of these sensorimotor tests have an acceptable reliability and could therefore be chosen with confidence to test the progress of a SMT intervention.

METHOD: 15 healthy recreationally active subjects aged between 18 and 25 years (5m / 10f) have twice undertaken the following sensorimotor tests with 6 weeks between the tests. During these 6 weeks, the subjects were not allowed to undergo any SMT. The subjects were allowed to continue their normal training program, however, were requested to fill in a training log.

Single-leg stance: The displacement of the CoP was measured by a force plate (Kistler, Winterthur, Switzerland) with a sampling frequency of 2000 Hz for 5 seconds per trial. The subjects had to stand on their dominant foot (the one they use to shoot when playing soccer) with arms akimbo and closed eyes. The mean values of the better two trials were measured for the path length and the moving area. The path length was calculated by summing up the distances between consecutive data points. The moving area was calculated as the summed up areas of the triangles between the geometric center of all the points and two consecutive data points.

Single-leg jump landing: At a single-leg jump landing from 36cm height the vertical, medio-lateral and anterior-posterior ground reaction forces were measured with a force plate (Kistler, Winterthur, Switzerland). The TTS scores were calculated with the sequential estimation method using an algorithm to calculate a cumulative average of the data points in a series by successively adding one point at a time (For more details see Ross et al. (Ross et al., 2005), Gribble et al. (Gribble and Robinson, 2009) or Wikstrom et al. (Wikstrom et al., 2005)). In addition, the medio-lateral displacement of the knee has been measured by filming the frontal plane movement of a cross that has been marked on the knee of the subjects. These videos have then been evaluated using a video tracking software (Skill Spector, video4coach, Svendborg, Denmark) to determine the maximal medio-lateral displacement of the knee. For all the single-leg jump measurements the mean of the best two out of three jumps has been taken for statistical analysis.

MFT S3-Check: For the MFT S3-Check the subjects tried to stay as calm as possible on a MFT platform for 30 seconds with their arms akimbo. The mean values of the better two of the three trials were taken for statistical analysis. The stability-index values are automatically calculated, whereas a low number stands for good balance.

Counter Movement Jump: Five CMJ with arms akimbo were performed by the subjects on a QuattroJump-platform (Kistler, Winterthur, Switzerland). The mean of the best two jump heights and the corresponding maximal RFD over 50ms were measured.

Leg-press: On a blocked 45°-leg-press with a fixed force plate on the foot-part, the subjects had to push as explosively and hard as they could. Their knee-angle has been set to 90° and the test has been repeated 5 times. The maximum isometric force and the maximal RFD over 50ms have been measured.

With Matlab R2009b the mathematical process has been executed and then the statistical analysis has been performed with SPSS Statistics 17.0. A dependent t-test for paired samples was applied to calculate, whether significant differences had occurred between the tests. With interclass-correlation the ICC-value has been calculated.

RESULTS: There are significant ($p<0.05$) differences between the pre- and post-test for the anterior-posterior TTS score and the medio-lateral knee displacement at single-leg landing (Tab. 1). Additionally there are trends towards significance for the vertical TTS score and the stability-index of the MFT S3-Check. All of these tests except for the stability-index, which shows moderate correlation, also show low correlation values. In addition also the medio-lateral TTS at single-leg landing and the RFD at CMJ show low correlation values. Both path length as well as motion area that were assessed at single-leg stance, show moderate to good correlation values. Good correlation values were achieved for the CMJ jump height, as well as for the measurements that were applied at the blocked leg-press, the maximal force as well as the RFD value. All subjects had filled out their diaries and abided to the demands not to undergo any SMT during the intervention.

Table 1. Mean values, number of subjects (N), standard deviation (SD), correlation (ICC) and level of significance (*p*) of the different sensorimotor tests, * significant ($p<0.05$).

	Mean	N	SD	ICC	*p*
Single-Leg-Stance_Path_pre [mm]	382.4	15	74.2	.775	.660
Single-Leg-Stance_Path_post [mm]	376.2	15	83.0		
Single-Leg-Stance_Area_pre [mm^2]	3906	15	1130	.859	.860
Single-Leg-Stance_Area_post [mm^2]	3936	15	1304		
Landing_Knee-Motion_pre [cm]	6.6	15	3.2	.208	.010*
Landing_Knee-Motion_post [cm]	4.4	15	0.9		
Landing_TTS_medio-lateral_pre [s]	1.44	15	0.08	.537	.111
Landing_TTS_medio-lateral_post [s]	1.41	15	0.06		
Landing_TTS_anterior-posterior_pre [s]	1.38	15	0.04	.441	.025*
Landing_TTS_anterior-posterior_post [s]	1.35	15	0.03		
Landing_TTS_vertical_pre [s]	1.32	15	0.07	.564	.091
Landing_TTS_vertical_post [s]	1.30	15	0.06		
CMJ_height_pre [cm]	41.1	15	7.4	.952	.837
CMJ_height_post [cm]	40.9	15	7.6		
CMJ_RFD50_pre [N]	451	15	159	.461	.711
CMJ_RFD50_post [N]	466	15	141		
Leg-Press_RFD50 _pre [N]	700	15	392	.824	.157
Leg-Press_RFD50 _post [N]	611	15	370		
Leg-Press_maxF _pre [N]	2100	15	735	.970	.332
Leg-Press_maxF _post [N]	2011	15	627		
MFT_Stability_pre	5.11	15	0.61	.695	.077
MFT_Stability_post	4.90	15	0.45		

DISCUSSION: Because of the good outcome values and the simple implementation, the single-leg stance is thought to be a robust method to test the static balance. A suboptimal choice to quantify improvements in balance or sensorimotor skills is the single-leg jump landing as correlation scores are very low, probably caused by the complexity of the movement in which slight changes in the motion sequence can lead to big differences in the measured scores. This outcome correlates with the findings of Ross et al. (Ross et al., 2005) who showed low to moderate reliabilities at TTS-scores in single-leg jump landings. The significant change in the values anterior-posterior TTS score and the medio-lateral knee displacement at single-leg landing could be explained by a learning effect due to the first test session. The jump height at CMJ seems to be a very useful and easily applicable dynamic method as long as no strength training is performed during the same intervention period as this can lead to falsification of the results. The dynamic RFD, however, is a very poor indicator as a slightly changed knee angle leads to a totally different RFD. This is also reflected in the low correlation value. Good correlation values were also found at the blocked leg press for the maximal force as well as the RFD. However, the same problem as with the jump height of the CMJ occurs: any strength training performed during an intervention period can lead to falsification of the results. As near-significant improvements have been found, the stability-index of the MFT S3-Check needs further investigation whether it is appropriate as a testing device for scientific studies or not. The movement is very task-specific and can probably be learned by training on a tilt-board. Based on the present study it is not possible to decide if a test is suitable to detect a change due to SMT, however it provides valuable data on the test reliability and therefore sets a baseline to judge whether differences from post to pre intervention may be accounted to the SMT or are in the range of interest variability.

CONCLUSION: Tests with a simple movement like the CMJ, the isometric RFD and the maximal force at the locked leg-press or the path length and motion area at single-leg stance show the highest reliability values. Specific movements (like S3) show a learning effect and at complex movements like the single-leg landing task there was almost no correlation, possibly because slight changes in the motion sequence lead to big differences in the measured scores. Therefore, such specific tests should not be applied to quantify general improvements of sensorimotor skills.

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MUSCULAR SYNERGISM DURING CORE STABILITY EXERCISES

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The aim of this study was to compare the myoelectric activity and synergism of Core region muscles among exercises commonly prescribed for Core training. The myoelectric activity of seven men was collected and the activation ratio among lumbar erector spinae (LES), lumbar multifidus (MT), external oblique (EO) and rectus abdominis (RA) were compared among eight exercises. The results suggest that EO has higher activation during frontal bridge, side bridge and "bird dog" exercises, RA has higher activation during frontal and side bridge, while LES and MT demonstrated higher activation during "bird dog" and double leg and single leg back bridge. We concluded that to train all muscles groups in a synergic way, in different postures, it should be prescribed at least one variation of the exercises that presents the flexor, lateral flexor and extensor pattern.

KEYWORDS: Biomechanics, Motor Behavior, EMG, Muscular Synergism, Core Training.

INTRODUCTION: The Core region functions as a "muscle belt" that stabilizes the lumbo-pelvic region, with or without the presence of upper and/or lower limbs movements (Kavcic et al., 2004). Besides being responsible for the lumbar and thoracic spine stabilization, it allows mobility and more efficacious upper and lower limbs force production and transfer (Akuthota & Nadler, 2004). It also acts as the center of the biokinematic chain in most daily and sport activities. The training of this region has been adopted by the community in order to increase athletic performance, as well as for clinical purposes in order to prevent and rehabilitate orthopedic injuries (Nadler et al., 2002, Tse et al., 2005, McGill & Karpowicz, 2009).
The term "stabilization exercise" has been used to denote any form of exercise that challenges stability of the spine, while muscle recruitment patterns, static and dynamic postures are trained (Akuthota & Nadler, 2004). The active stability is provided mainly through co-contraction of muscles present in this region to alleviate the overload on the trunk (McGill, 2007). Despite the importance in understanding the strategies of trunk stabilization provided by these muscles, Kavcic et al. (2004) reported that few studies have assessed the activation and muscle co-contraction during different core stability exercises. Thus, the purpose of this study was to compare the myoelectric activity and synergism of Core region muscles among exercises commonly prescribed for strengthening and stabilizing core region.

METHODS: Seven men experienced in strength and core training (body mass: 78.8 ± 10.5kg, height: 180.3 ± 7.8cm, age: 27 ± 6 years) participated in the study. The myoelectric activity of Lumbar Erector Spinae (LES), Multifidus (MT), External Oblique (EO) and Rectus Abdominis (RA) was collected (BIOPAC Systems Inc., California) during the execution of eight core stability exercises commonly prescribed (Table 1). Each exercise was performed in randomized order for 30 seconds with 5 minutes rest among them. The signals were filtered by a fourth order Butterworth filter, with cutoff frequencies of 20Hz and 400Hz. RMS values were obtained, at each 5ms signal, for 10 seconds (range 5 to 15 seconds) and normalized by the greatest RMS value obtained during two maximal voluntary isometric contractions (MVIC), in positions of flexion, for RA, extension to the LES and MT, and lateral flexion of the trunk, for the EO. The muscle synergism between couples of muscles was calculated by dividing the value of normalized myoelectric activation in the following conditions: EO/RA, EO/LES, EO/MT, RA/LES, RA/MT, LES/MT. The values of synergism and values of each individual muscle were compared among exercises through the Friedman's non parametric test of analysis of variance with repeated measures and Dunn's post-hoc tests. The level of significance was set at 5%.

Table 1. Name, legend and figure of the exercises used in this study.

Exercise	Frontal Bridge	Right Side Bridge	Left Side Bridge	"Bird Dog"*	"Bird Dog"**	Right Single Leg Back Bridge	Left Single Leg Back Bridge	Double Leg Back Brigde
Legend	FB	RSB	LSB	BDRA	BDLA	RBB	LBB	DBB

* Performed with right arm and left leg lifted;
** Performed with left arm and right leg lifted;

RESULTS: The statistical test detected differences in the EO/RA synergism (p = 0.0002) and post-hoc test showed that these differences occurred between LSB and BDLA, RBB and BDLA, DBB and BDLA. For the EO/LES synergism statistical test also detected significant differences among exercises (p <0.0001), and the post hoc test showed differences between FB and RSB, FB and DBB, RSB and DBB. The analysis of variance detected differences in EO/MT synergism (p <0.0001) and post hoc test revealed that these differences were present between the exercises FB and RBB, FB and LBB, FB and DBB, RSB and DBB, LSB and DBB, BDLA and DBB. In relation to the synergism of RA with LES (RA/LES) and MT (RA/MT) significant differences were identified (p <0.0001). The post hoc test identified that the significant differences between the exercises were similar in both situations (FB and LBB, FB and DBB, RSB and DBB, LSB and DBB) with the exception of the condition FB and BDLA, which were not identified differences in the RA/MT synergism. For the LES/MT synergism were not found statistically significant differences (p = 0.3704). Analyzing the individual muscles, all of them showed statistical significant differences. For the EO the Dunn's test detected differences between the situations RSB and FB, DBB and FB, RBB and RSB, RSB and DBB, BDLA and DBB (p <0.0001). For the RA (p <0.0001) post hoc test showed differences in the situations FB and LBB, FB and DBB, RSB and LSB, RSB and DBB. For LES (p = 0.0002), the post hoc test identified differences between the situations FB and RSB, FB and RBB, FB and BDLA, LSB and RBB. MT (p <0.0001) showed differences between the situations FB and RSB, FB and BDLA, FB and LBB, RSB and LSB (Table 2).

Table 2. Values, described as mean (standard error), of the synergism in the six situations and individual myoelectric activity of the four muscles in each exercise. In the right column the p values are presented. Similar symbols below each row represent significant differences between two exercises.

	FB	RSB	LSB	BDRA	BDLA	RBB	LBB	DBB	P value
EO (%)	65.5 (16) * &	52.5 (11.5) * % ^	9.5 (2.6)	9.6 (3.3)	36.6 (6.9) #	6.0 (1.6) %	10.5 (2.7)	2.6 (0.6) & ^ #	<0.0001
RA (%)	33.5 (7.8) * &	40.0 (13.5) # ^	10.8 (2.8) #	5.5 (1.2)	6.0 (1.7)	8.1 (3.2)	5.7 (2.0) *	3.5 (1.1) & ^	<0.0001
LES (%)	5.0 (1.3) * & #	37.2 (5.5) *	7.5 (1.3) ^	31.1 (6.5)	38.6 (6.2) #	42.1 (7.9) & ^	34.7 (2.3)	33.1 (4.1)	0.0003
MT (%)	4.2 (0.6) * & #	37.8 (7.5) * %	6.5 (0.7) %	19.5 (4.1)	35.9 (7.5) &	24.7 (3.5)	30.9 (4.5) #	25.4 (2.9)	<0.0001
EO/RA	2.3	2.9	1.3	2.3	7.9	1.1	2.4	1.1	0.0003

	(0.7)	(1.5)	(0.7) *	(1.1)	(2.3) * & %	(0.3) &	(0.8)	(0.5) %	
EO/LES	12.7 (6.5) * &	1.4 (0.3) * %	1.0 (0.3)	0.3 (0.1)	1.0 (0.2)	0.1 (0.0)	0.3 (0.1)	0.1 (0.0) & %	<0.0001
EO/MT	15.6 (2.4) * & %	1.4 (0.1) #	1.3 (0.3) $	0.5 (0.1)	1.1 (0.2) ^	0.2 (0.1) *	0.3 (0.1) &	0.1 (0.0) ^% #$	<0.0001
RA/LES	7.5 (3) * & %	1.6 (0.7) #	1.6 (0.3)	0.2 (0.1)	0.2 (0.1) %	0.2 (0.1)	0.2 (0.1) *^	0.1 (0.0) & # ^	<0.0001
RA/MT	8.7 (1.7) * &	1.1 (0.3) #	1.6 (0.4)	0.3 (0.1)	0.2 (0.0)	0.3 (0.1)	0.2 (0.0) * ^	0.1 (0.0) & # ^	<0.0001
LES/MT	1.8 (0.4)	1.1 (0.2)	1.2 (0.2)	1.6 (0.2)	1.2 (0.2)	1.7 (0.3)	1.2 (0.2)	1.3 (0.2)	0.3704

DISCUSSION: In this study we compared the myoelectric activity and muscle synergism among various abdominal and trunk extensors exercises and our findings are comparable and similar to the study of Kavcic et al. (2004) with respect to the main exercises that activate trunk flexors and extensors. The exercises FB, RSB, LSB demonstrated a flexor pattern, while the exercises BDRA, BDLA, RBB, LBB and DBB showed an extensor pattern (Kavcic et al., 2004).

During the BDLA it was verified that the EO has a significant activation (36.6%), which demonstrates the importance of this muscle during asymmetric activities. We postulate that these results are related to the oblique pennation angle of the fibers of this muscle, which was ratified by the small actions of RA, which has a more longitudinal angle of pennation (Oatis, 2008). The differences between the EO/RA synergism occurred in the exercises that there are a predominance of the extensors muscles, which was expected since the RA is an antagonistic to these muscles and EO is not. Only in the LBB and BDRA, characterized by an extensor pattern, we did not find differences due to the low recruitment of EO.

The RSB was that exercise that required high muscle activity of both flexors and extensors of the right side, however, LSB did not demand high activity of these muscles. This suggests that these exercises have an unilateral dominance of muscle recruitment that could be an important ally in the correction of asymmetries of strength and power between the right and left sides (McGill et al. 1999).

Although the analysis of muscle synergy is essential for the initial understanding of the strategies of lumbar stabilization, the study of the activity of individual muscles is also significant because the lumbar stability is achieved with low percentages of activation, about 20% to 40% of maximum (McGill, 2007).

The RSB and FB were the exercises that required greater activation of the abdominal muscles, reaching percentages from 50% to 65% for EO and 30% to 40% for RA. Thus, we believe that training these exercises should be recommended only for experienced individuals, since the high muscle activation can modify the stability of lumbar spine leading to premature fatigue. The recommendation would be that for untrained individuals such exercises are initiated with the knee on the ground, to decrease the lever arm. The training of the muscle EO, should be preceded by bird dog exercise prior to such exercises (RSB and FB), since all the active muscles remained within the recommended levels (around 30% of maximum) for training the lumbo-pelvic stability (McGill, 2007).

In the exercises RBB and LBB the muscle activity of MT and LES remained within the recommended range, but the participants reported a high rate of subjective effort, especially in the gluteus and back thigh regions. Therefore, it would be advisable to prescribe bilateral exercises (DBB) initially, since the activity of lumbar muscles is similar regardless the

number of mechanical constraints. The exercises BDRA and BDLA were active within the recommended range.
Future studies with the inclusion of exercises on unstable surfaces and in other populations, as untrained beginners, could be very valuable to complement the data obtained in this study.

CONCLUSION: We conclude that in order to train all muscle groups synergistically in different postures at least one variation of exercises that show a flexor pattern (FB, RSB), a variation of exercises that presents a lateral flexor pattern (FB, RSB and BALA) and exercises that have an extensor pattern (RBB, LBB and DBB) should be performed. Among the possible variations, it could be included a decrease of the lever arm of the exercises, for example, with the knee on the ground during frontal and lateral bridge. The aim of these changes is to develop a pedagogical progression, which allows the inclusion of exercises with greater ease in the early stages of training. This would allow adaptation to training stimuli in a less harmful manner to the lumbar spine.

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INTEGRATING SPORT BIOMECHANICS AND EXERCISE PHYSIOLOGY FOR TRAINING COLLEGIATE ATHLETES DURING A COMPETITION SEASON

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KEYWORDS: sport specificity training, volleyball, strength and conditioning

INTRODUCTION: Sport specificity training involves the design and implementation of strength training and conditioning programs tailored to a specific sport with the goal of optimizing performance. When training collegiate athletes during a competition season there is an even greater emphasis placed on optimization. This is because the NCAA places time restrictions on collegiate athletes that effectively limit the volume of training that any one athlete can participate in per week. As a result, coaching staffs need to maximize time devoted to skills based training while still ensuring sufficient time is allocated for developing an athlete to peak physical conditioning and rest (Marques, et al., 2006).
The goal of this paper is to present a framework for integrating sport biomechanics and exercise physiology within the design of sport specific training programs. Research from both fields has direct applicability to sport specificity training and integration is often an implicit dimension to such research. Yet, comparably few studies have been explicit about how to best integrate biomechanics and exercise physiology within the context of sport specificity training. Although this paper specifically uses Women's Collegiate Volleyball as a case study, the intent is to initiate discussion regarding the need to explicitly integrate sport biomechanics and exercise physiology when developing strength training and conditioning programs for collegiate athletes.

METHODS: First, a typical, mid-week, in season, Women's Division II Collegiate Volleyball practice was filmed. This practice was broken down by drill to determine three different components. The first component was to determine the biomechanics of sport specific movements performed by a starting outside hitter, middle hitter, setter, and libero. Player movements were counted whenever a sudden change of player position on the court was identified during the work portion of a drill. These movements were broken down according to cardinal directions with north being oriented towards the net. Additionally, three sport specific movements were included. These included the attack jump, the block jump, and defensive crouch. The second component was to determine the relationship of on-court movements to strength and conditioning exercises. These were determined by the authors who have multiple years of experience designing strength and conditioning programs for collegiate volleyball. Moreover these lifts are commonly integrated into current strength and conditioning programming for volleyball (Hedrick, 2007). Lastly, the volume and intensity of on-court movements and foot contacts were quantified by counting the number of movements identified above and calculating the work-to-rest ratio. The work-to-rest ratio was calculated by using an internal clock embedded into the video media software (Windows Media Player, 2007). Work time was calculated starting at the beginning of play/drill and stopped at the end of play determined by a dead ball or an abrupt end of player activity (Iosia & Bishop, 2008). Using data from the film analysis a periodized, sport specific training program was written.

RESULTS AND DISCUSSION: Table 1 summarizes the analysis of the film data for a starting outside hitter and indicates that several sport specific movements common to volleyball are identifiable. Each of these movements seek to maximize lower body and upper body power, which are intended to rapidly change an athlete's horizontal and vertical position in the playing area. For instance, when conducting the attack jump the athlete engages in two phases of movement to generate maximum vertical jump height along with forward momentum toward the net before transitioning into a third phase of movement intended to generate upper body power to attack the volleyball (Cisar & Corbelli, 1989).

Translating the biomechanics of sport specific movements to strength training and conditioning exercises should consider the different phases of a single action, in this case the attack jump. Plyometrics such as box jumps are a good example of a common strength and conditioning exercise that corresponds only to the first two phases of the attack jump. However, from a biomechanics perspective the attack jump is a multi-planar, multi-jointed action, as are most movements specific to the sport of volleyball. In order to replicate the transition into the third phase of the attack jump variations of Olympic-style lifts such as a fast tempo front squat transitioning into a jerk or push-press may be more appropriate. Variants of lifts such as these more closely reflect the multi-joint, multi-planar movement specific to the sport, rather than focus solely on one muscle group in a single plane like the squat alone does (McGill, et al., 2009). More so, such lifts develop both speed and strength, which are the two components of power.

Prescribing volumes based on number of repetitions for a periodized strength training and conditioning program that is sport specific also benefit from data collected through an analysis of practice film. For instance, within the microcycle it will be necessary to shift to a lower volume during the course of the week. The goal of this reduction is to provide a taper for the athlete to maximize performance in weekend matches. Thus, early in the week it may be beneficial to overload the athlete with exercises that correspond to on-court movements identified in the film study. However, in order to taper for competition a reduction of volume should occur. This is because, as the film data shows, a substantial volume is already being reached during team practices. The danger of prescribing additional workouts during the taper phase of the microcycle is that it may result in overtraining and decreased performance. Similarly, quantifying the volume of in practice movements using film, measured as total number of foot contacts, can provide an additional source of information for determining overall training loads prescribed throughout the competition season macrocycle. The goal at this scale is, again, to avoid overtraining athletes while ensuring that they continue to make gains in physical conditioning in order to peak at the onset of post-season play.

Table 1. Summary of film data for a starting outside hitter during a typical mid-week competition season practice.

Movement	Muscles Recruited	Example of corresponding exercise	Volume
Attack Jump	Primary: Trunk, lower extremity, shoulder and upper extremity	Tempo front squat (or power clean) to jerk	53
Block Jump	Primary: Lower extremity, trunk. Secondary: upper extremity	Tempo front squat (or power clean) to push press	23
Defensive Crouch	Primary: Lower extremity, trunk	Half squat	22
Lateral movement	Primary: Lower extremity, trunk	Ground based Plyometrics	394

CONCLUSION: Sport biomechanics and exercise physiology play a significant role in designing sport specific training programs for collegiate athletes. Explicitly integrating research from these two fields is one way to optimize an athlete's physical conditioning in light of limitations placed on training schedules during the competition season. The framework advocated here suggests that sport specific movements occurring during organized team practices should be factored into the design of periodized, sport specific, strength training and conditioning programs. This framework integrates sport biomechanics with exercise physiology and employs the use of film analysis as one method to analyze sport specific movements and quantify the volume of those movements during practices.

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ISBS 2010

Oral Session 17

Gymnastics

TECHNIQUES TO START THE STOOP CIRCLE (ADLER) ON HIGH BAR

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The stoop circle rearward forward (Adler) is often performed by gymnasts on high bar. The final position, the handstand is a component of judges' execution value. For the stoop circle only the final position is determined by the gymnastics rules. The start position is not declared. There are two main techniques to start the element stoop circle, which we define as "high" and "low" technique. Aims of the research are finding biomechanical differences between the two techniques and deducing requirements for gymnasts. 2D-video analysis from high bar routines of the World Championships 2007 was used. There are more gymnasts performing the "high" technique. We find differences in movement time and maximum angular velocities for hip and shoulder angles. These differences should be considered by coaches and gymnasts.

KEYWORDS: artistic gymnastics, high bar, stoop circle, hip angle, shoulder angle.

INTRODUCTION: The stoop circle rearward forward, also called "Adler" (german word for "eagle") is an old element in artistic gymnastics. It is performed mostly by men on high bar, but also by women on the uneven bars. In the last years this element becomes more important because of the possibility to combine it with another element, especially with flight elements. This combination is important to earn combination points to get a higher difficulty value of the routine. There are different variations for the stoop circle (see Figure 1). Gymnasts perform the element without a turn, with half turn or full turn but always through the handstand. The handstand is a good position for the next flight elements.

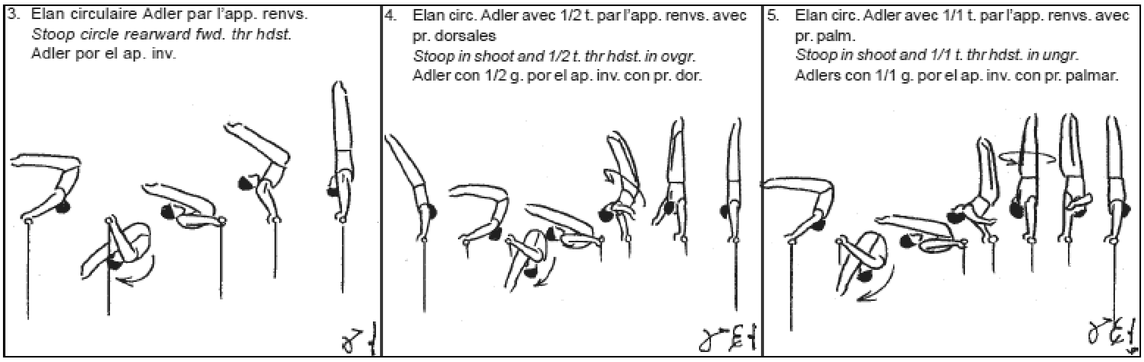

Figure 1. Different variations of the stoop circle in high bar in the Code of points (Féderation Internationale de Gymnastique, 2009, 130).

For the stoop circle (with or without turns) there are two main techniques for the first part of the element (Figure 2 and 3). The first possibility is to start the stoop circle from handstand. The gymnasts finish their giant swing forward and after they reached the vertical line over the high bar (handstand position with open shoulder and hip angle) they stoop in. For the second technique the gymnasts do not finish their giant swing. Shortly after they pass the horizontal line on high bar they bend their shoulder and their hip angle for stoop in. For a better understanding we declare the first technique (stoop in from handstand) as the "high" technique (high centre of mass, CM) and the

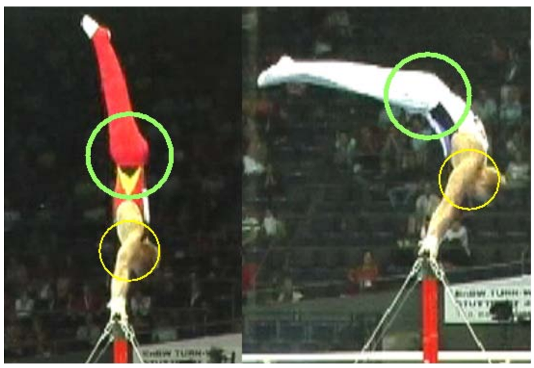

Figure 2. Characteristics to define the two techniques by hip and shoulder angle higher or lower than 160° (left: "high" technique, right: "low" technique).

second technique (stoop in after horizontal) as the “low” technique (lower CM). Using a biomechanical description we separate the stoop circle depending on the hip and shoulder angle at the moment when the CM passing the vertical over the high bar (Figure 2). If the hip and shoulder angles were higher than 160° we have the “high” technique and if the hip and shoulder angles were lower than 160° we define this as the “low” technique.

Figure 3. The two possibilities to start the stoop circle rearward forward (top: “high” technique, bottom: “low” technique).

For the stoop circle rearward forward there are no judge’s rules for the start position. The Code of points (Féderation Internationale de Gymnastique, 2009) and other regulations from the gymnastics Federation (Stoica, 2009) specify only the last position (handstand) of the element. If the gymnast did not reach the handstand position and there is a difference of more than 15° the execution judges deduct 0.1, 0.3 or 0.5 points. Additionally the level of difficulty will be downgraded by the difficulty judges (e.g. from C-value to B-value or no value).

For coaches the questions are “Which technique enable better progression to more advanced skills?”, “Which technique is better to reach the perfect final position (handstand)?” and “Which requirements are important for the different techniques?”. Using a biomechanical approach we can help the coaches to answer especially the last question.

There are a lot of publications concerning different elements on high bar (dismounts, e.g. Hiley & Yeadon, 2003; 2008 and flight elements e.g. Hiley, Yeadon & Buxton, 2007), but there was no paper found for the stoop circle.

METHOD: We recorded 160 men’s high bar routines at the Artistic Gymnastics World Championships 2007 in Stuttgart in all parts of the competition (qualification and finals). A descriptive frequency analysis of the different techniques was the first step. A two-dimensional video analysis (fixed DV-Camera Panasonic NV-GS 300, 50 Hz, 2D-DLT) was utilized for detailed analysis. We digitized 7 body landmarks (ankle, knee, hip, shoulder, head, elbow and wrist) from 15 stoop circles and calculated the CM (Saziorski, Aruin & Selujanow, 1984) and the hip and shoulder angle (smoothed data by cubic spline function). To compare the two techniques we divide the stoop circle in 4 phases, which are defined by the position of the CM related to the high bar (Figure 4).

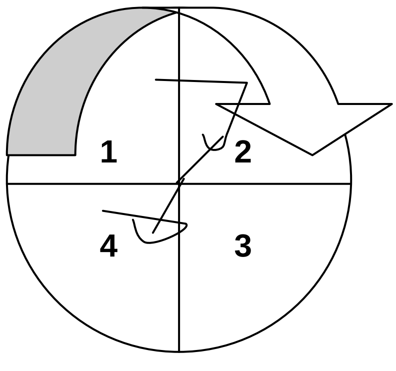

Figure 4. Phases of the stoop circle.

- Phase 1: CM from left horizontal to upper vertical
- Phase 2: CM from upper vertical to right horizontal
- Phase 3: CM from right horizontal to lower vertical
- Phase 4: CM from lower vertical to left horizontal

If the movement is executed counter clockwise left and right horizontal is switched. After the 4th phase the gymnast start the turn for stoop circle with half or full turn.

RESULTS: From 160 men's high bar routines 100 include one ore more stoop circles. 57 percent of all stoop circles were performed with the "high" technique. Only for stoop circles with half turn the frequency is near equal (Table 1).

Table 1. Frequency of stoop circles

Variation of stoop circle	"High" Technique	"Low" Technique
Without turn to handstand	34	26
With half turn to handstand	24	25
With full turn to handstand	21	8
Sum of all variations	79	59

If we compare the movement time for the two techniques for the four phases the "high" technique has longer first and second phase but shorter Phase 3 and 4 (Table 2).

Table 2. Mean time [s] for the phases of stoop circles

Phase	"High" Technique	"Low" Technique
1	0,81 (±0,11)	0,64 (±0,08)
2	0,58 (±0,14)	0,43 (±0,05)
3	0,24 (±0,06)	0,33 (±0,19)
4	0,30 (±0,12)	0,33 (±0,16)

Table 3. Mean maximum velocities [°/s] for the phases of stoop circles

	"High" Technique		"Low" Technique	
Phase	shoulder	hip	shoulder	hip
1	218 (±54)	171 (±52)	238 (±39)	246 (±69)
2	406 (±85)	506 (±104)	322 (±69)	486 (±233)
3	333 (±86)	168 (±64)	298 (±29)	141 (±46)
4	430 (±115)	369 (±107)	483 (±66)	320 (±114)

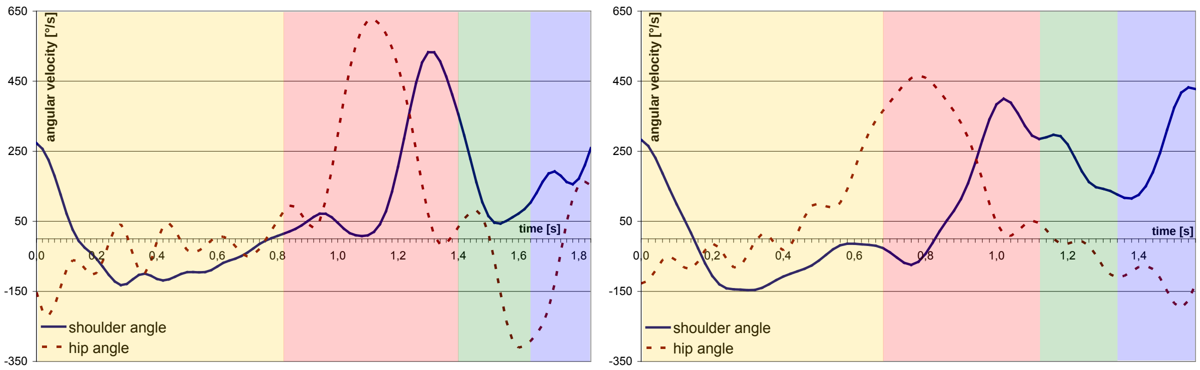

Figure 5. Examples for angular velocities of "high" (left) and "low" (right) Techniques of the stoop circles (phases separated by the coloured background).

For the coaches especially the velocities of hip and shoulder angle were interesting. In the downwards Phases (2 and 3) the absolute maximum angular velocities of hip and shoulder angle are higher for the “high” technique (Table 3). The examples in Figure 5 should illustrate these findings. The gymnasts with the “high” technique (left) start his movement of hip and shoulder angle later but must do it faster.

DISCUSSION: More gymnasts prefer the “high” technique with the late stoop in. But our findings show this technique make higher demands on the gymnasts. They must bend their hip and shoulder faster than with the “low” technique. Our first results give us no answer why gymnasts use the “high” technique with greater demands. One assumption of the coaches was that body mass and height of the gymnasts is one reason for choosing “high” or “low” technique. Using official information from the gymnastics federation with height and mass of the gymnast we could not find differences between the gymnasts using “high” or “low” technique.
Maybe the question is not only answerable with biomechanical data. It could be also a question of which movement has the greater effect on judges and spectators. This is a question of aesthetics. Maybe it is more impressively to have a slow start of the stoop circles and than a faster movement (“high” technique) compared to the slower movement with an early start (“low” technique).

CONCLUSION: Both techniques of the stoop circle rearward forward were performed successful by many gymnasts. But our analysis with kinematic data shows different requirements in hip and shoulder angle for these techniques. Using the “high” technique higher angular velocities must be performed in hip and shoulder angles. Coaches and gymnasts should consider this in their training especially in strength training for hip and shoulder angle. Further research should include the calculation of energy and joint torques. Applying these data to a simulation model could be used for movement optimization.

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BIOMECHANICAL CHARACTERISTICS OF WHOLE-BODY FAST REACHING MOVEMENTS

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Executing a whole-body fast reaching movement is an essential skill in many competitive sports. The present study investigated this kind of movement by biomechanical analyses. Five male university athletes and five fencing team members volunteered as subjects in this study. Each subject was asked to perform whole-body fast reaching movements and end at a stable posture. A motion capture system was used to record the kinematic data. The results demonstrate that the fencers moved the upper extremities earlier than the lower extremities, and the proximal segments started slightly before the distal segments. In conclusion, in executing whole-body fast reaching movements, upper and lower extremities work together as one functional unit, and there may be slightly different strategies adopted by each individual according to their exercise habits.

KEYWORDS: functional synergies, coordination, movement pattern

INTRODUCTION: Although reaching a specified target is required in many tasks, the choice of the trajectory between the initial and the final position remains free (Alexander, 1996). The elbow and shoulder joints have been shown to link together as a single unit when pointing or reaching (Kaminski, 1995). In many athletic competitions, performing whole-body fast reaching movements is even more essential. In this kind of movements, not only the upper extremities but also the lower extremities and the trunk need to participate in the task. At higher speeds, the more pronounced initial posture adjustment and straighter hand path may be necessary to maintain equilibrium because of conflicting dynamic balance and performance constraints (Pozzo, 1998).

When executing whole body reaching movements, a particular synergy among moving joints has been found previously with the focus on upper extremities. We hypothesized that when the target position is fixed, there is a specific sequence of moving body segments for each person. The purpose of this study is to clarify the coordination strategy among moving joints in whole-body fast reaching movements.

METHOD: Ten male university athletes volunteered as subjects in this study, and the subjects are separated into two groups. Group 1 consisted of common male university athletes (subject 1 to subject 5) specialized in different sports, and group 2 consisted of fencing team members (subject 6 to subject 10). Subjects warmed up and practiced the reaching movement before data collection. All reaches started with a standard stance (with right foot ahead of the left one 40% of the leg length), right shoulder in 0^0 of abduction, right elbow in 90^0 of flexion, and left palm on the anterior superior iliac spine. Each subject was asked to perform whole-body fast reaching movements and remain stable after finishing the motion. Rather than reaching for a small target, a large area was used to exclude the possibility of reducing reaching speed for higher accuracy. Each subject had to complete ten trails with barefoot. The stopping posture is specified with right arm straightforward, right knee in 90^0 of flexion, trunk leaning forward and in line with the straight left leg, and both heels remaining on the ground.

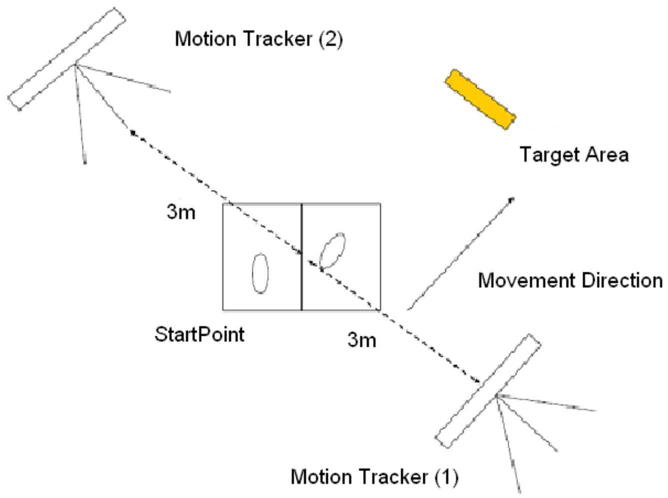

Figure 1. The set up of the experimental condition.

Two Visualeyez motion trackers with the sampling rate of 150 Hz were positioned on both sides of the subjects to record movements. Markers were placed along each side of the body on the acromion, elbow, wrist, sacrum, greater trochanter, femoral epicondyle, lateral malleolus, heel, and the fifth metatarsal. The kinematic data were collected by the software VZSoft, and then filtered by Matlab 7.0 (4th-order butterworth filter with low-pass frequency at 6Hz).

RESULTS: Table 1 shows the timing (% of movement completion) when the maximum angular velocity occurred in each joint in the fastest motion. Table 2 shows the starting timing of extension angular velocity of each joint in the fastest movement. The results demonstrate that the fencers moved the upper extremities earlier than the lower extremities, and the starting timing of maximum angular velocity of the proximal segments (upper arm and thigh) occurred earlier than the distal segments (forearm and shank).

Table 1. Timing of maximum angular velocity

	R_elbow(%)	R_shoulder(%)	R_knee(%)	R_ankle(%)	L_knee(%)	Velocity(m/s)	Time(s)
Group 1	75.55*	74.45*	66.58	69.44	66.57	1.52	0.62
(n=5)	(3.12)	(3.55)	(10.16)	(8.23)	(5.69)	(0.18)	(0.04)
Group 2	52.60*	55.15*	63.41	67.19	73.54	1.42	0.69
(n=5)	(11.47)	(15.10)	(8.13)	(10.03)	(8.99)	(0.17)	(0.09)

The symbol * indicates the significance level p<.05 between the 2 groups.

Table 2. Starting timing of maximum angular velocity

	R_elbow(%)	R_shoulder(%)	R_knee(%)	R_ankle(%)	L_knee(%)	Velocity(m/s)	Time(s)
Group 1	37.36*	29.92*	31.56	36.10	32.77	1.52	0.62
(n=5)	(5.86)	(3.44)	(5.13)	(8.22)	(4.58)	(0.18)	(0.04)
Group 2	13.36*	15.80*	32.05	38.52	41.80	1.42	0.69
(n=5)	(8.15)	(7.42)	(10.41)	(18.10)	(10.88)	(0.17)	(0.09)

The symbol * indicates the significance level p<.05 between the 2 groups.

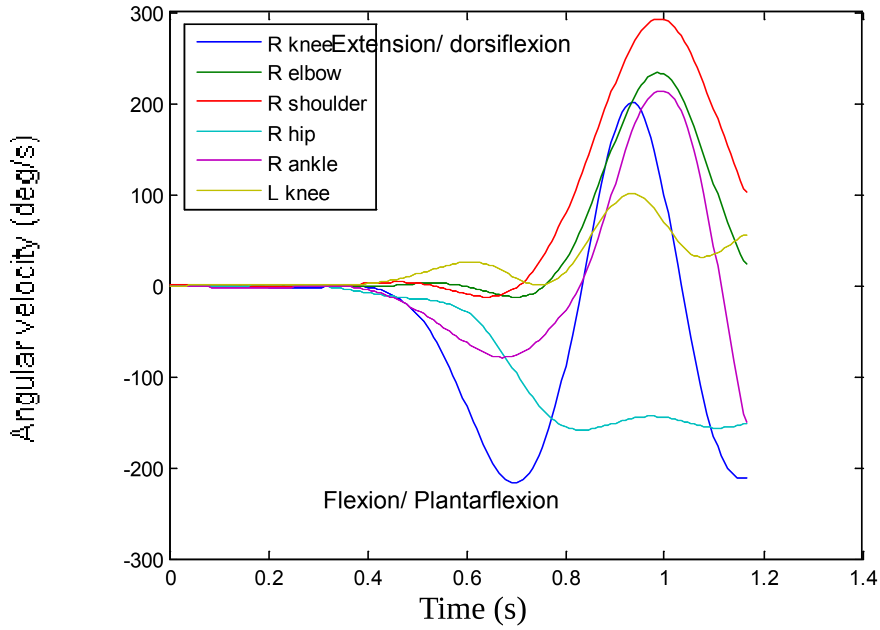

Figure 2. Angular velocity of subject 1.

DISCUSSION: The purpose of the present study is to clarify the coordination strategy among moving joints in whole-body fast reaching movements. In similar studies, Kaminski et al. (1995) showed that elbow and shoulder joints are linked together as a single unit. Kaminski (2007) indicated that the reach and postural synergies became coupled, resulting in the arms, legs and trunk working together as one functional unit to move the whole body forward in reaching motions.

In this study, the maximum angular velocity occurred roughly at the same time of the acromion, elbow, knee, and ankle joints of right side. Most of subjects' maximum angular velocity of left knee occurred around the same time as the other joints. This indicates that upper and lower extremities work together as one functional unit in the motion, similar to the results of previous studies.

The proximal segments started slightly before the distal segments, which conformed to the kinetic chain theory. The fencing members activated the upper extremities earlier than the lower extremities. The reason is that they be training to always move their arm first in order to make their swords hit the opponent as fast as possible. This indicates that exercise habits may strongly affect movement patterns.

CONCLUSION: The results of this current study provide a certain coordination pattern of whole-body fast reaching movements. That is, in executing this kind of movements, upper and lower extremities work together as one functional unit, and the proximal segments started before the distal segments, and there may be slightly different strategies adopted by each individual according to their exercise habits.

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KINEMATIC CHANGES DURING LEARNING THE LONGSWING ON HIGH BAR

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Understanding technique development during complex skill learning provides information that can be used to influence feedback and skill development. The purpose of this study was to investigate changes in longswing technique during an 8 week period of learning. Fourteen male participants with no previous high bar experience took part in the training study. Data were collected using a CODA motion analysis system (200 Hz) during weekly testing sessions. There was a significant increase in swing amplitude for the group between week 1 and all subsequent weeks ($p < .05$). Based on initial swing amplitude three patterns of learning were displayed; each having distinctive functional phase characteristics. This study highlights the importance of quantifying changes in technique throughout learning on an intra-individual basis, to understand how technique changes.

KEYWORDS: Technique, skill acquisition, functional phase

INTRODUCTION: Previous literature has reported differences between novice and expert technique during gross complex motor skills (Delignières *et al.,* 1998), however, few have considered the nature of how novices' technique changes during a period of learning. Understanding how technique develops during learning provides precise information that can be used to influence feedback.

In men's gymnastics the longswing on high bar is a key skill which underpins the development of more complex skills. The biomechanics of performing successful longswings are well understood. Research has emphasised the importance of movements at the hip and shoulders, specifically, a hyper-extension to flexion action of the hips and hyper flexion to extension action of the shoulders occurring beneath the lower vertical (Arampatzis & Brüggemann, 1999; Yeadon & Hiley, 2000; Irwin & Kerwin, 2005). Irwin & Kerwin (2007) termed these movements Functional Phases (FP) since 70% of the gymnast's musculoskeletal work was found to occur during this part of the skill. Knowledge of the biomechanics of successful longswings, specifically the FPs, provides a theoretical underpinning to address more applied issues associated with learning longswing technique. Initial insights into kinematics associated with learning the longswing have been provided by Busquets *et al.* (2009) who described changes in coordination between the hip and shoulder for a novice cohort after a two month practice period in comparison to that of experienced gymnasts, and Williams *et al.* (2009) who provided a comparison of FP characteristics between performers at different stages of learning. However, the specifics of how longswing technique changes during a period of learning are not known. The purpose of the current study was to investigate the position of FPs of the 'looped bar longswing' performed by novices over a period of learning, in order to identify key kinematics of technique associated with learning.

METHOD: 14 male participants with no prior high bar experience (age 20 ± 3 years, mass 73 ± 7 kg, height 1.76 ± 0.06 m), volunteered to take part in this study. The participants consented to learn the 'looped bar longswing' (LLS), a mechanically similar but safer variant of the traditional 'chalked bar longswing' (Irwin & Kerwin, 2005). The longitudinal study comprised an initial testing session in which participants were shown videos and received an explanation of the aims of the LLS before attempting the skill. Testing sessions required each participant to perform 5 trials of 3 swings with the ongoing aim of increasing swing amplitude. A gymnastics coach provided support to assist each participant in gaining initial angular momentum. Data were collected during each trial for each performer. The testing sessions were interspersed with training sessions throughout the study. During training sessions, longswing specific skills and conditioning exercises reflective of those used in contemporary coaching environment were performed in a gymnasium.

Unilateral kinematic data were collected using an automated 3D motion capture system (CODA) sampling at 200 Hz. Two CX1 CODA scanners (Charnwood Dynamics Ltd, UK) provided a field of view exceeding 2.5 m around the centre of the bar. Active markers were placed on the lateral aspect of each participant's right side at the estimated centre of rotation of the shoulder and the elbow, mid forearm, greater trochanter femoral condyle, lateral malleolus, fifth metatarsophalageal and the centre of the underside of the bar. For individuals, measures of height and mass were obtained, digital images facilitated the calculation of all other anthropometric data for use with a geometric inertia model (Yeadon, 1990) to obtain individual-specific body segment inertia parameters.
Swing 2 in each trial was analysed, ensuring a full independent attempt was being performed. Circle angle (θ_C) was defined by the mass centre to bar vector with respect to the horizontal. In order to provide inter-performer comparisons of swings, data were interpolated in 1 degree increments of rotation about the bar. Lines joining the shoulder centre, greater trochanter and femoral condyle markers defined the hip angle (θ_H). Shoulder angle (θ_S) was defined by the lines joining elbow, shoulder and greater trochanter markers. Hip and shoulder angles (θ_H; θ_S) were differentiated to create angular velocity (ω_H; ω_S) profiles. 2D coordinate data were processed with the kernel smooth function (MathCad14™) with the smoothing parameter set to s = 0.10.
The performance measure; Swing amplitude (θ_{CA}), was defined as the circle angle between maximum height of the mass centre on the downswing to maximum height on the upswing. FP analysis of the hips was described by the position of maximum hyper-extension (θ_{CH1}) to flexion (θ_{CH2}) in θ_C, and shoulders by maximum hyper flexion (θ_{CS1}) to extension (θ_{CS2}) in θ_C. Differences across testing sessions were quantified using repeated measures ANOVA. Statistical significance was set at $p < .05$. Maulchy's test was used to determine the sphericity assumption within the data; where sphericity was violated probability was corrected according to the Greenhouse-Geisser procedure. Post hoc comparisons were made on the resultant data. Bonferroni corrections were applied for multiple comparisons.

RESULTS: Group mean θ_{CA} showed significant increases between session 1 (178 ± 42°) and session 8 (322 ± 41°), p <.05. The most rapid improvements occurred between the first and second testing sessions p <.05 (Fig. 1.).

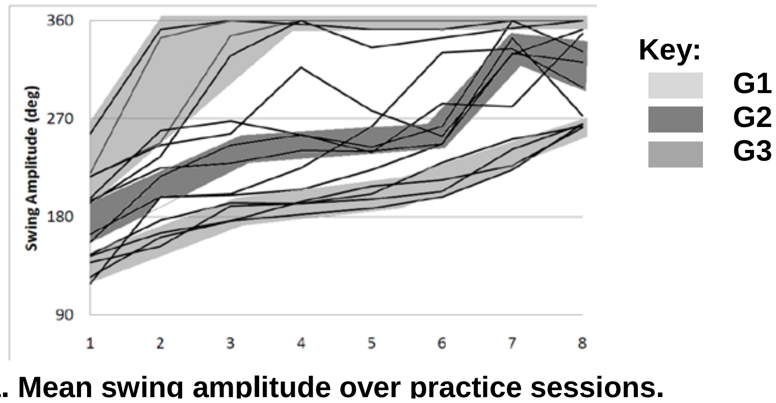

Figure 1. Mean swing amplitude over practice sessions.

Increases in θ_{CA} followed there distinct trends (Fig. 1). Based on these trends, three groups were defined; G1, G2, G3 as follows:
G1 *High learning rate* (n=4): participants were able to perform the skill by session 3. Largest amplitude swings were demonstrated by these participants. (Fig. 1). FP analysis identified that θ_{CH1} became progressively later during the learning period (p< .05). During successful LLS in sessions 3-8, θ_{CS1} occurred earlier and θ_{CS2} later than in sessions 1 and 2 (Table 1, Fig. 2).
G2 *Variable learning rate* (n = 5): demonstrated an initial increase in θ_{CA} between weeks 1 and 3, plateaued within the middle weeks before increasing during the penultimate week (Fig. 1). There were significant changes in hip FP variables θ_{CH1} and θ_{CH2} (p< .05), but not in the corresponding shoulder variables, θ_{CS1} and θ_{CS2}.

G3 *Low learning rate* (n = 4): participants began and ended with the smallest swing amplitude but demonstrated a steady increase over the 8 sessions (Fig. 1). No significant differences occurred in θ_{CH1} during the 8 weeks, but θ_{CH2} advanced between sessions 1 and 8 (p< .05). After session 2, the onset of the shoulder FP (θ_{CS1}) did not change (Fig. 2).

Table 1. Session mean results for representative participants from G1, G2, G3 of the onset and termination of functional phase of the hips (θ_{CH1}, θ_{CH2}) and shoulders (θ_{CS1}, θ_{CS2})

	G1				G2				G3			
Session	θ_{CH1}	θ_{CH2}	θ_{CS1}	θ_{CS2}	θ_{CH1}	θ_{CH2}	θ_{CS1}	θ_{CS2}	θ_{CH1}	θ_{CH2}	θ_{CS1}	θ_{CS2}
1	185	290	186	368	230	325	241	346	202	288	285	320
	(8)	(17)	(10)	(12)	(12)	(11)	(16)	(15)	(10)	(12)	(9)	(4)
2	177	278	182	355	214	332	206	371	209	288	209	322
	(10)	(12)	(6)	(22)	(3)	(13)	(11)	(14)	(3)	(4)	(10)	(20)
3	222	338	150	381	191	313	181	385	204	297	202	335
	(60)	(23)	(8)	(8)	(4)	(18)	(11)	(10)	(4)	(5)	(4)	(14)
4	236	345	151	384	192	321	194	343	211	304	209	358
	(38)	(4)	(11)	(7)	(6)	(23)	(31)	(56)	(6)	(10)	(6)	(8)
5	251	350	163	392	191	298	175	346	206	296	202	354
	(9)	(4)	(10)	(8)	(9)	(22)	(13)	(56)	(4)	(7)	(4)	(8)
6	260	348	149	387	192	335	171	393	200	300	197	352
	(12)	(9)	(3)	(8)	(7)	(23)	(17)	(10)	(18)	(8)	(18)	(8)
7	260	356	166	398	156	349	227	403	190	308	202	361
	(11)	(5)	(4)	(7)	(31)	(20)	(68)	(17)	(16)	(17)	(33)	(8)
8	250	351	150	392	203	346	211	330	207	335	200	371
	(13)	(1)	(13)	(14)	(11)	(18)	(44)	(55)	(5)	(8)	(31)	(9)

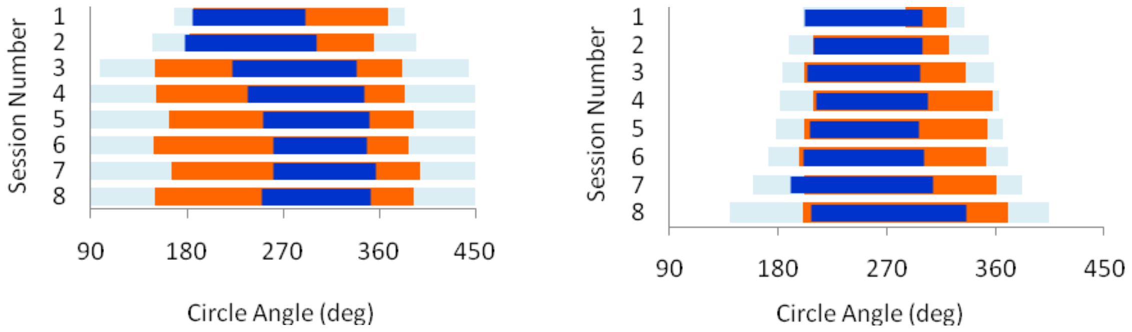

Figure 2. Mean functional phases (FP) of the hips (blue line) and shoulder (orange line) during the circle angle (light blue) for G1 (left) and G3 (right)

DISCUSSION: The aim of this study was to describe changes in the position of FPs of the LLS performed by a group of novices during a period of learning. Swing amplitude (θ_{CA}) significantly increased during the 8 week learning period. Three patterns of change in θ_{CA} (Group 1-3) were evident based on 2 criteria; the initial θ_{CA} and subsequent pattern change. Specifically, those participants who obtained the greatest θ_{CA} during session 1 were able to successfully perform the LLS by session 3 (G1). Four participants who had a mid-range θ_{CA} showed inconsistent increases in θ_{CA} over sessions (G2). Those who began with the smallest θ_{CA} in session 1 had the smallest θ_{CA} during session 8, sustaining a steady increase in θ_{CA} over sessions (G3). Therefore, the nature of improvement in the current skill appears to be closely related to skill level during initial attempts. Evidence from contemporary motor learning literature indicates that learning rate is individual and task specific even when a persistent change is apparent across subjects (Newell *et al.*, 2001). During learning a lateral swinging task on a suspended platform, Teulier & Delignières (2007) found between participant differences in initial coordination pattern complexity, however during subsequent trials it was reported that technique evolved in a similar manner. Though, during this study initial θ_{CA} appears to be a good predictor of final success of the LLS. For similar tasks (Teulier & Delignières, 2007) it has been suggested that, based on Newell's (1986) categories of 'individual-specific organismic constraints' differences in initial skill levels can be related to organismic factors, as task and environmental constraints were constant for all participants.

FP analysis revealed individual specific changes throughout the learning period; analysis of a representative performer from G1, G2 and G3 were presented. Results showed differences in the ability to adjust the placement of the FP within the θ_C across sessions. Specifically, G1, appeared able to significantly change the onset and termination of the hip and shoulder FP throughout learning. G3 appeared to adjust hip FP, but not shoulder FP. In contrast, G2, were unable to significantly change the onset of the FP after session 2. Newell *et al.* (1989) suggested that variability in movement patterns permits the exploration of a motor-perceptual workspace, and was therefore an inherent characteristic of functional dynamical systems when learning a given motor task. For example, the more successful participants appear to be able to alter placement of both the hip and shoulder FPs in order to create a movement pattern that enabled them to increase θ_{CA}, however it appeared that less successful participants were unable to alter these aspects of FPs in order to improve performance. Based on theories of motor learning, these findings have potential implications for the types of training intervention provided to a novice performer learning the LLS. For example, it could be interpreted that differences in initial θ_{CA} are related to organismic constraints of the system, where it is the ability to vary the onset and termination of the hip and shoulder FP which enable a progression to successful LLS. As such, it could be suggested that for a less successful performer (G3), skill progressions which promote actions temporally similar to that of the LLS for which firstly the hips (as per G2), and then the hips and shoulder actions (as per G1) would aid a performer in changing from an initially inadequate motor behaviour (Newell *et al.,* 2001).

CONCLUSION: This study highlights the importance of quantifying changes in technique throughout learning, and on an intra-individual basis when seeking to investigate the nature of technique modifications. These experimental findings could provide key information which is required when considering motor learning in a sports context, in order provide the most effective feedback to a performer during skill acquisition. Additional work is required to explore if a relationship exists between the placement of the FP and θ_{CA} via kinetics analysis.

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Oral Session 18

Aquatics

COMPUTATION OF HIP AND SHOULDER TORQUES IN COMPETITIVE SWIMMING

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The mechanisms of propulsion can be deduced from 3D video analysis of swimming. The basic hydrodynamic equation can be used to compute the total force for those particles. The shoulder torque was calculated by summation over hand, lower arm and upper arm of infinitesimal torques of displaced water particles. These muscle force moments were related to the velocity of the mass centre as a measure of the propulsion. However, a direct interrelation between velocity and torques cannot be established since the total resistance of the body in motion is unknown. Therefore, the aim was to determine individual differences in swimming technique, controlled by shoulder, hip torques and swimming velocity, at any state of one movement cycle. Recommendations for best propulsion techniques are derived.

KEYWORDS: 3D analysis, hydrodynamics, swimming, muscle force moment, propulsion.

INTRODUCTION: For more than 40 years researchers have invested much work into a better understanding of propulsion in swimming. Propulsion is reflected by the velocity of the mass centre, which can be calculated by 3D video analysis. For training methodology it is interesting to know in which way propulsion is related to swimming technique. To understand this relation the drag forces acting on the limbs as well as the total drag acting on the body have to be known. This is a difficult problem. The paper presents the results of a six year study which consisted of the following elements: Design of a feasible measuring system, determination of the limb velocity by 3D video analysis, explanation of propulsion from a phenomenological point of view based on measuring data, quantification of the net muscle force moments by using the basic hydrodynamic equation (1) and, finally, deducing recommendations for the improvement of the individually best propulsion techniques. Our main aim was to indicate individual differences in swimming technique depending on hip and shoulder torques and on swimming velocity.

The calculation of forces is based on 22 body point coordinates and their derivatives. There are three different forces in fluids: inertial force, convective force and – in contrast to solid bodies – also pressure force. In international scientific literature there are no publications paying attention to this distinction of three forces; application of the basic hydrodynamic equation for the calculation of propulsion forces has yet to be used. Sanders (1997), for example, applied the following formula to compute the force on the hand $\mathbf{F} = \rho\, A\, (C_X, C_Y, C_Z)\cdot\mathbf{v}^2 + \rho\, A\, (D_X, D_Y, D_Z)\cdot\mathbf{a}$, where **C** is a constant referring to drag and lift and **D** is a vector including the effective mass accelerated by the hand with acceleration **a**. The acceleration term is quite similar to the first term in the basic hydrodynamic equation, the velocity term however is different. Explanations for movement in the water run from the mechanics of the paddle steamer (Schleihauf, 1979), or that of the ship's propeller up to undulations, like vibrations of the fins of fishes. Currently the vortex theory by Matsuuchi et al. (2009) is most popular. To move quickly in water we use our relatively large hands and feet to find water resistance and we repel ourselves like from a soft solid. In crawl stroke this happens directly with the hands when they are moved against the swimming direction. When swimming crawl stroke or butterfly with high speed, no part of the legs can push the water against the swimming direction. The feet are moved forward all of the time (Hildebrand, 2001). In the downstroke of the legs the streaming water (viewed from the swimmer) generates an acceleration force. Since the legs are bent in the knee and the hip joints, the muscle force moments cause a stretching of the body, pushing the mass centre while the joints are opening. The upstroke is even more complicated. The feet are coming up close to the surface. One can see the formation of a water peak at the highest point of the feet. Muscle forces cause a lift of the legs which in turn induces a negative drag. In other words, if the upstroke is strong enough the inflowing water sliding upwards along the legs will be accelerated. By Bernoulli's equation, a slipstream arises at the lower legs, which can be transformed into propulsion at the beginning of the downward movement. This tricky technique must be learned.

METHOD: The experiment was carried out at the swimming flume at the Olympic Training Centre in Hamburg. Technology and software were elaborated by Drenk, Hildebrand, Kindler and Kliche (1999) at the Institute for Applied Training Science Leipzig. Two 50 Hz cameras were used. We recorded the performances of four elite swimmers of height 182.0 ± 3.5 cm and of age 23.8 ± 4.3 years; one male breaststroke swimmer and three female swimmers. We calculated the forces at the limbs and moreover the joint torques at elbow, shoulder, knee and hip joint. The flow speed ranged from 1.15 m/s to 1.65 m/s. From 3D analyses we obtain velocities and accelerations of all model points. Using velocities, accelerations and coordinates of the joints by linear interpolation we calculate the velocity distribution of arms and legs (Drenk et al., 1999).

The basic hydrodynamic equation involving force $\mathbf{F}$ on a volume element and its velocity $\mathbf{v}$ is

$$\mathbf{F} = \rho\, \partial\mathbf{v}/\partial t + \rho\, (\mathbf{v} \text{ grad})\, \mathbf{v} + \text{grad } p, \quad (1)$$

where p denotes the pressure, ρ the fluid density $\rho\, \partial\mathbf{v}/\partial t$ the inertial force and $\rho\, (\mathbf{v} \text{ grad})\mathbf{v}$ the convective force. Explicitly, this reads as follows

$$\mathbf{F}_1 = \rho\, (\partial u/\partial t + u\cdot\partial u/\partial x + v\cdot\partial u/\partial y + w\cdot\partial u/\partial z) + \partial p/\partial x$$
$$\mathbf{F}_2 = \rho\, (\partial v/\partial t + u\cdot\partial v/\partial x + v\cdot\partial v/\partial y + w\cdot\partial v/\partial z) + \partial p/\partial y$$
$$\mathbf{F}_3 = \rho\, (\partial w/\partial t + u\cdot\partial w/\partial x + v\cdot\partial w/\partial y + w\cdot\partial w/\partial z) + \partial p/\partial z.$$

Since velocity distribution and pressure close to a moving body and other boundary conditions are not known, the boundary value problem cannot be solved.
We use these equations as a definition of the force. We do not consider the whole swimmer, but only limbs which produce propulsion. Thus we can assume that water particle displacement occurs only close to the propulsive areas, and there are no shearing forces. The hands move on an S-shaped curve in a way that the hands catch into stationary water all the time. The resulting velocity, acceleration and velocity gradient are put into the right hand side of the basic hydrodynamic equation. On the left hand side we then obtain the force acting on a single mass element. Finally we can compute the torques via cross product of force and distance vector to the joints. We apply the following three simplifying assumptions:

1. To calculate the force at the edge of limbs we take the effect of the vortices into account by designing an experimental form factor equal to two (Sommerfeld, 1954).

2. Instead of the hydrodynamic pressure we use the hydrostatic pressure $p = p_o + \rho\, g\, z$, z being the depth of the water, p_o the atmospheric pressure and g the gravitational acceleration.

3. The unknown velocity at some inner point of the limb is the linear interpolation of the velocities at the end points. The velocity in a small area close to the limb is constant in any plane orthogonal to the vector from one endpoint to the other. The same applies to accelerations.

These simplifications refer to the unknown local environment, the area close to the hand, shank and so on. The force $\mathbf{F}$ related to the mass unit has three terms: the partial time derivative $\rho\, \partial\mathbf{v}/\partial t$, the inner product $\rho\, (\mathbf{v} \text{ grad})\, \mathbf{v}$ and the gradient of the pressure grad p. The partial time derivative is the acceleration of the replaced particle. To compute the second term $(\mathbf{v} \text{ grad})\, \mathbf{v}$, including $u\cdot\partial u/\partial x$, $v\cdot\partial u/\partial y$, $w\cdot\partial w/\partial z$, we apply assumption 3: the gradient of velocity is constant and multiple of the endpoint vector. Here we also take the above mentioned form factor two from assumption 1 into account. In the first approximation the convective force is proportional to the square of the velocity of the moved body segments. Net muscle force moments for shoulder and hip joints can be calculated from the resulting force on mass particles. This is done by taking cross products of forces and distance vectors to the joints and summation over all mass particles.

RESULTS: The principle which is used to produce propulsion with arms is a little bit different from the principle for legs. In breaststroke initially, after diving into the water, the arms move in swimming direction and against the water. The water is flowing upward towards the shoulder. When the arms are taken to the body there is one moment when the inner palm, the forearms and parts of the upper arms are pushing against swimming direction on the unmoved water resulting in a reverse of the streaming direction of the water. Obviously, just in these phases the drag force can be completely transformed into propulsion. It does not matter if the hands are led exclusively against the swimming direction, since the resulting force is always created against the local streaming direction. In the crawl stroke the hands are mainly led backward towards the hip. Especially in breaststroke swimming they are also led laterally. The drag that the hands are faced with is used with the help of the muscle force moment in shoulders and arms in such a way that the trunk pulls forward (Hildebrand, Drenk, & Kliche, 1999). Referring to crawl stroke one could imagine a hold at an anchor in the water from which one is pushing off. In breaststroke swimming one is pushing the hands together against the imaginary anchor to pull the body forward over the shoulders. But in these two cases different muscle groups are working (Fig. 2). The principles are implemented individually: In case the stretched arm catches deeply into the water a big torque is resulting from the long hand-lever (Fig. 1, left). In case the forearm is quickly moved into a perpendicular position towards the swimming direction drag is created at the whole forearm. This force lasts longer (Fig. 1, right).

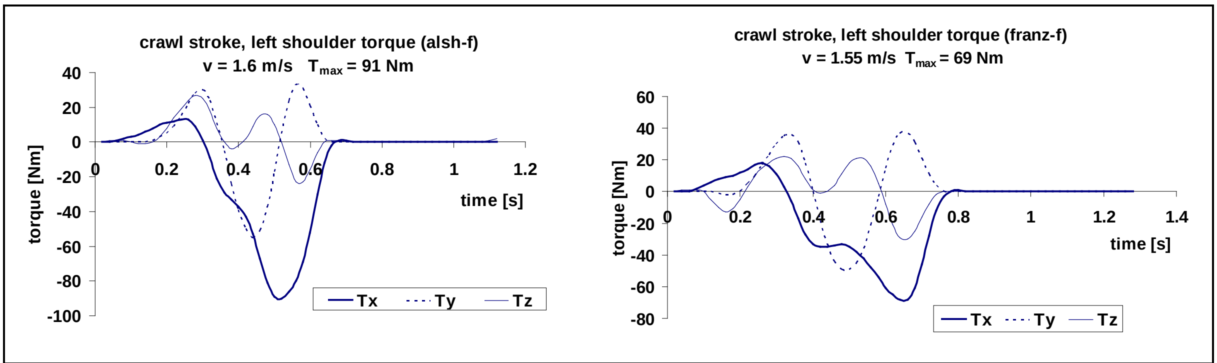

Figure 1. Left shoulder torque of two elite crawl stroke swimmers. Left figure: deep arm pull, v-flow = 1.6 m/s, Right figure: perpendicular lower and upper arm and pressure on forearm until the arm is leaving the water, v-flow = 1.55 m/s.

The thick solid line T_x represents the shoulder torque around the transverse axis, on the left with a maximum of 91 Nm and on the right with a maximum of 69 Nm. The dotted line T_y represents the torque component around the longitudinal axis and finally the thin solid line T_z with the smallest amplitude represents the component around the vertical axis. The technique shown on the left picture requires both higher force values and higher joint performances. These two swimming techniques require different dry land training.

In breaststroke the movement of arms differs between women and men. Men swim about 0.5 m/s faster than women. For men, by lack of time, a backward movement of the hands against the water is impossible. The very first technique of women breaststroke, in which hands create pressure against the swimming direction, becomes more and more ineffective with increasing swimming speed. Men develop the following strategy: when the outward movement of the hands is completed they are immediately put together below the breast. This move generates a drag transversal to the swimming direction.

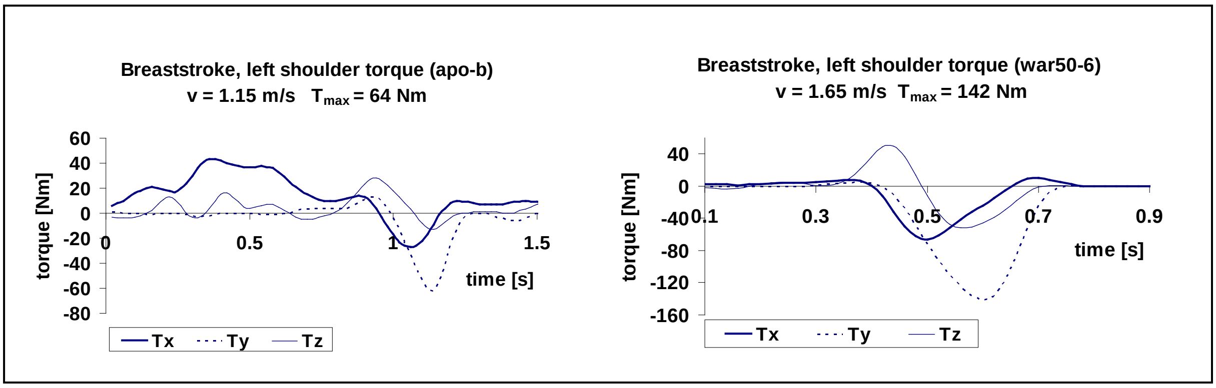

Figure 2. Breaststroke, left arm torques. Left figure: swimming technique at v-flow = 1.15 m/s. Right figure: swimming technique at v-flow = 1.65 m/s.

The result is a much greater torque about the body axis, as shown in Figure 2 (dotted line, right figure). This technique therefore requires different control of arm and shoulder muscles, and a training that is different from the one for the classical technique.

CONCLUSION: The quantification of the individual propulsion moments improved our understanding of the propulsion processes. We can prove that there are different ways of creating propulsion depending on swimming velocity. This has to be taken into account especially in long and short distance breaststroke swimming as well as when learning swimming techniques. As expected, the contribution of the moments arising from the upper arm and the thigh are small compared to the moments resulting from hand, foot and lower leg movements. In between we found individually different maximum forearm moments. At this point the swimming technique can be optimized. Compared to earlier estimations (Hildebrand, 2001) the torque in the shoulder joint is 50 percent larger. The significance of the three terms in the basic equation is of great interest. The greatest gain is produced by the convective force, while the effect of the pressure gradient is significantly smaller (that could affirm assumption 2). Inertial force has only the same significance for long breaststroke distances. In breaststroke swimming, with a velocity of 1.60 m/s, arm propulsion is dominating, but when swimming with about 1 m/s, leg propulsion dominates. The dolphin stroke in butterfly swimming and crawl stroke swimming (in case it is performed) has almost the double effect compared to the typical breaststroke leg stroke. Thus it has been possible to prove the great importance of the dolphin stroke for the performance trend in swimming.

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THE CONSISTENCY OF FORCE AND MOVEMENT VARIABLES AS AN INDICATOR OF ROWING PERFORMANCE

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A portable biomechanical collection system was used to test fourteen male, coxless pair rowing crews under simulated race speeds. The data was used to examine if the consistency of 8 biomechanical variables, calculated using a modified Coefficient of Variation (CoV), were related to the overall performance, in terms of velocity, of the crews. It was found that 6 of the variables demonstrated a significant correlation to overall boat speed, with the consistency of Normal Gate Force (r=0.737), Handle Velocity (r=0.758) and Trunk Velocity (r=.757) showing very strong correlation. It was concluded that while not imperative to the outcome, consistency of force application and movement patterns may be important in rowing performance.

KEYWORDS: rowing, force, consistency, movement

INTRODUCTION: In an effort to gain a better understanding of performance outcomes in the sport of rowing, researchers have steadily looked beyond the variables able to be derived through the examination of force production and related factors such as work and power. Whilst acknowledging that these factors have a large influence on performance, it has increasingly been realised that certain other mechanisms within the rowing stroke may be used in an effort to discriminate between skill levels (Dal Monte and Komor, 1988).
The notion of consistency of output has been examined as a discriminator in several studies. Smith and Spinks (1995) examined several work output and skill based variables derived from an ergometer test, in an effort to discriminate between subjects of different rowing ability, with stroke-to-stroke consistency of the handle force being found to be a significant indicator of classification. In a similar on-water study, it was again found that the consistency of force production was an important variable for discriminating between competition levels (Smith et al., 1994).
To further investigate this method as a potential discriminator of performance it was decided to extend the scope of the consistency quantification from the singular force applied to the handle, to an examination of four relevant force variables and four movement variables. The forces examined were the two forces measured at the gate, being those normal and transverse to the face of the gate, and the horizontal force applied to the left and right sides of the foot stretcher. The movement variables examined were the velocities of the 3 main components of the rowing stroke, namely the seat velocity, which equates to speed of leg drive, the trunk extension/flexion velocity measured as relative speed of C7 to the seat, and the velocity of the arms. A final movement variable was also included, being the handle velocity, which may be considered the overall product of the combination of the 3 previous components.

METHOD: The participants in this study consisted of 7 lightweight and 7 heavyweight male, coxless pair crews. The mean height and weight of the rowers was 187.1 ± 6.2 cm and 81.7 ± 10.3 kg respectively. All participants were currently training with an Australian sports institute and data collection was carried out as part of routine biomechanical testing. Each crew were tested whilst rowing in their preferred boat, which was equipped with a portable biomechanical data collection system. The system utilised purpose designed force transducers contained within specially designed gates and foot stretchers which enabled the measurement of two-dimensional gate force and the horizontal force applied to the left and right sides of the foot stretcher.

Linear displacement of the moving seat and approximate position of C7 was also collected using drum and reel potentiometer transducers, similar to those previously used (Kleshnev, 2000; Smith and Loschner, 2002). The location of C7 was used to determine the position of the top of each athlete’s trunk in reference to relative seat movement, providing an indication of trunk flexion and extension throughout the stroke. It is a limitation of the study that trunk movement could not be measured more precisely. Velocity of the arms was calculated as the difference in velocity between the oar handle and the C7 point. A similar arrangement has previously been used to estimate the proportion of segmental contributions to overall velocity and power (Kleshnev, 2000). Boat velocity and acceleration were determined by a Rover unit (James et al., 2004), which utilised a 100 Hz accelerometer coupled with a 1 Hz GPS receiver to determine instantaneous boat velocity. All data were sampled at 100 Hz and transmitted via radio telemetry to a receiver attached to a laptop via USB. Data was collected to hard drive as testing progressed, and was displayed on the laptop screen as numerical values for all the variables.

Prior to data collection, each crew was required to perform a pre-race warm-up to their satisfaction, followed by a step rate piece over a 2000m rowing course. At the completion of the 2000 m the participants were then required to return to the start of the last 500 m of the rowing course and, when they felt prepared, perform one 500 m trial at a full race effort, during which data was collected for this investigation. Athletes were instructed to perform the 500m at what they would consider their “mid-race” tempo and speed.

Post collection, 15 consecutive stokes were selected from each trial piece for further examination. This is a common method of obtaining representative ensemble data in biomechanical testing of rowing (Smith and Draper, 2006). The strokes were separated into single strokes using the position of zero degrees of the strokes oar during recovery, i.e., when the oar was perpendicular to the long axis of the boat. Each stroke was then time normalised to 101 points and variability of each parameter was calculated using a modified version of the Coefficient of Variation (CoV). The calculation of the modified CoV is shown, and expressed as a percentage, with 100 % representing repeated curves which are identical.

$$CoV = 100 \cdot \left(1 - \frac{\sqrt{\frac{1}{n}\sum \sigma_i^2}}{\frac{1}{n}\sum |X_i|} \right)$$

where:
n = number of intervals in the cycle
σ_i = st. dev. of variable X at ith interval
X_i = mean of variable at ith interval

Signal filtering was applied only to those variables that would be used to derive secondary variables (i.e. oar angle deriving oar speed) in order to minimise the introduction of noise due to differentiation. The cut-off frequency applied to the Butterworth low pass filter was determined by plotting the filter residual error versus the smoothing frequency (Winter, 1990) over a range of cut-off frequencies.

Bivariate correlations were performed to establish the relationship between the primary performance indicator of average boat velocity and the CoV of the four force and four movement variables. Level of significance was set at 0.05.

RESULTS AND DISCUSSION: The mean stroke rate of the race piece was 33.8 ± 1.0 strokes per minute. Results of the bivariate correlations are presented in Table 1.

Table 1. Calculated CoV of biomechanical variables and correlations to average boat velocity

		Normal gate force	Transverse gate force	Left stretcher force	Right stretcher force	Seat Velocity	Handle Velocity	Arm Velocity	Trunk Velocity
CoV(S.D.)		90.4(2.5)	84.4(3.9)	82.3(4.8)	83.8(5.0)	91.7(2.1)	94.4(1.3)	88.5(2.5)	90.9(2.2)
Average velocity	r	0.737**	0.578*	0.429	0.402	.599*	.758**	.633*	.757**
	Sig	0.004	0.038	0.144	0.174	.031	.003	.020	.003

** Correlation is significant at the 0.01 level (2-tailed).
* Correlation is significant at the 0.05 level (2-tailed).

Examination of the correlations demonstrated that there was a significant correlation between the CoV of both normal gate force ($r = 0.737$, $p<0.01$) and transverse gate force ($r = 0.578$, $p<0.05$) to the average velocity of the boat. As the value of CoV can be affected by the magnitude of the mean value of the variable, and it has been previously shown that applied force may also be a good indicator of boat velocity (Leighton, 1983), it was necessary to conduct partial correlations between the variability of forces and boat speed, controlling for mean values of the force, the outcome of which are shown in Table 2

Table 2. Partial correlation matrix of variability controlling for mean force

		Normal gate force	Transverse gate force	Left stretcher force	Right stretcher force
Average velocity	r	0.649*	0.602*	0.454	0.114
	Sig	0.023	0.038	0.138	0.725

* Correlation is significant at the 0.05 level (2-tailed).

After taking into account the magnitude of the relevant forces it can be seen that the CoV of the force applied in a direction normal to the gate still exhibits a significant relationship to average velocity ($r=0.649$, $p<0.05$), as does that of the transverse force, or outward force applied along the long axis of the oar ($r=0.602$, $p<0.05$). Conversely there was no significant relationship found between the CoV of either the left or right foot stretcher force to the performance outcome of the boat.

The higher degree of variability of force applied to the foot stretcher may be attributed to a number of factors involved in the dynamics of rowing. During the recovery phase the only contact the athlete has with the boat is through the seat and foot stretcher. The recovery is also the least stable part of the stroke, with change in boat orientation, particularly roll of the boat, generally more noticeable (Loschner et al., 2000). While many athletes use the height of the oar handle during the recovery to stabilise this movement, some degree of balance is obtained through the adjustment of pressure on the feet. This must be performed while the seat is moving towards the stern, where force on the stretcher is used to halt momentum of the body in the latter stages of recovery. This effect is compounded in a crew boat, in which the rower is not only reacting to boat orientation but also the movement of his partner. It appears that while force on either foot is quite variable through the stroke, by the time the kinetic chain of the body is involved to transfer force developed at the stretcher to the handle, the variability in force is greatly reduced.

All four of the movement variables demonstrated a significant relationship to boat velocity, with the CoV of both the handle and trunk velocity exhibiting a correlation at the $p<0.01$ level ($r = 0.758$ and 0.757 respectively). The variability in arm velocity had a relationship of $r = 0.633$ ($p = 0.020$) and that of seat velocity $r = 0.599$ ($p = 0.031$). These results suggest that the manner in which crews perform the gross movement patterns within their technique may have a relationship to performance, in that those crews that more consistently repeated the same movement pattern stroke after stroke tended to exhibit higher average boat velocity over the same period. This may simply be due to the fact that the faster crews were more

proficient at performing these tasks, however, the crews were all of a similar standard in terms of rowing experience.
It appears that for this sample, the consistency of applied gate may be used as a discriminator in the prediction of performance outcome. This observation was extended to include a number of other variables, as significant correlation occurred between the major variable related to performance, that of average boat velocity, and the variability of forces applied to both the face and side of the gate, as well as the consistency of all 4 of the movement variables examined.

CONCLUSIONS

It appears that the use of variability of biomechanical variables in rowing may be of some use in predicting performance outcome. There was shown to be a significant relationship between the low variability in the force applied both normally and transversely to the face of the gate and the performance outcome of boat velocity. No significant relationship was evident for the foot stretcher force or within either the lightweights or heavyweights when treated as separate groups.
A similar trend was found when examining the consistency of the four selected movement variables of seat, body, trunk and handle velocity. Crews who exhibited a lower variability in these variables also tended to exhibit a higher average boat velocity. It is difficult to determine if reduced variation in these factors has a direct implication on performance outcome or, as outlined by Smith and Spinks (1995), are merely identifying better ‘skilled’ crews who also happen to be inherently faster.

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MEASURING THE WAVE DISSIPATION PRODUCED BY A SWIMMING-LINE SEPARATION ROPE

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KEYWORDS: Swimming, hydrodynamics, wave drag.

INTRODUCTION: Hydrodynamic drag (D) seems to be one of the major determinants of swimming performance. D is usually divided into pressure, friction and wave drag (D_w). Meanwhile, D_w can be due to two distinct phenomena: (i) wave production ($D_w$wp) and (ii) transfer of negative wave momentum ($D_w$tm). $D_w$wp refers to the energy dissipated from the kinetic energy of the swimmer and used to generate waves, and $D_w$tm refers to the drag effect (reduction of forward kinetic energy of the swimmer) attributed to the impact of waves produced by others, or produced by the swimmer itself and rebounded at a swimming pool wall. In order to define the competition lane of each swimmer, the competition swimming pools dispose of swimming-line separation ropes (S-LSR). In the meantime, the manufacturers of this S-LSR claim that they have the ability to absorb waving energy, and thus to dissipate waves avoiding $D_w$tm, and other perturbing wave effects. The purpose of this research was to characterize the swimmer's wave production, and to measure the effect upon the wave energy dissipation of a common S-LSR (Fig.1).

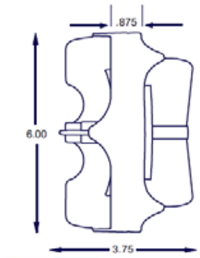

Figure 1. Profile of the S-LSR studied.

METHOD: Two experienced and trained competitive swimmers (25 / 17 years, 1.83 / 1.78 m, 83.8 / 72.1 kg, 56.02 / 59.61s 100 medley) performed 5 x 12 m in each competitive swimming technique (N = 10) with and without S-LSR, at maximal velocity, and at the mean maximal velocity obtained for breaststroke (1.3 m/s). During each 12 m swim, a 6 m section (between 4 and 10 m) was timed using a stopwatch and an experienced operator. The interval between repetitions was casuistic and decided based on the inexistence of dynamometric evidences of water disturbance. At a distance of 2 m from the swimming lane an extensiometric force-plate of 0.5x0.5 m (Pereira et al., 2006) was fixed on the wall of the swimming pool (Fig. 2).

The S-LSR was fixed at 1.15 m from the force-plate. The force-plate was operating at 1000 Hz, and data was exported to a PC using a 16 bit A/D Biopac converter and the Acqknowledge 3.2.5 software. Dynamometric data were acquired during 24 s in each trial, from where the most prominent wave was selected to extract variables corresponding to this wave-train. These include wave amplitude (difference, in N, from the lower to the higher value registered in one wave) and intensity (the highest positive value, in N, registered in each wave train, after removal of the offset). In each side of the S-LSR a vertical ruler was mounted, allowing to measure the height of the waves using the high-speed video mode (300 Hz) of one Casio-Exilim F1 camera. SPSS and ANOVA were used after checking for normality (Shapiro-Wilk).

Figure. 2. Picture of the experimental setup. The force-plate and the rulers are shown.

RESULTS: The main results of the study are presented in Table 1.

Table 1. Mean ± standard deviation of the dynamometric and wave height values

		Without Swimming-line separation rope				With Swimming-line separation rope			
		Velocity (m/s)	Wave intensity (N)	Wave amplitude (N)	Wave height (m)	Velocity (m/s)	Wave intensity (N)	Wave amplitude (N)	Wave height (m)
Maximal velocity	Crawl	1.7±0.8	40.4±7.8	66.8±15.9	0.079±0.025	1.7±0.5	28.3±8.0[a]	54.0±11.8[c]	0.047±0.022[a,c]
	Back	1.5±0.8	36.3±5.8	53.0±12.8[e]	0.050±0.010[e]	1.5±0.6	27.3±5.8[a,c]	50.5±10.6[c]	0.033±0.010[a,c]
	Butterfly	1.6±0.7	34.3±7.0	58.0±14.0	0.040±0.014[e]	1.6±0.9	26.2±5.2[a]	48.1±12.7	0.032±0.01[a]
	Breast	1.3±0.4	33.1±4.1	65.6±4.5	0.048±0.017	1.3±0.9	20.5±7.0[a]	36.5±13.9[a]	0.023±0.009[a]
Breaststroke mean velocity	Crawl	1.3±0.9	27.1±9.2[b]	46.9±19.2[b]	0.064±0.030[b]	1.3±0.6	15.9±3.9[a,b]	27.1±7.4[a,b]	0.026±0.006[a,b]
	Back	1.3±0.7	25.9±5.0[b,c,d]	43.4±6.2[c,d]	0.061±0.023[b]	1.3±0.5	22.6±3.4[e,d]	42.3±6.6[e,d]	0.038±0.007[a,c,e]
	Butterfly	1.3±0.9	38.4±13.1	69.2±14.3[e]	0.060±0.011[b]	1.3±0.8	34.2±3.2[b,c,e]	56.2±8.9[a,c,e]	0.045±0.015[a,b,c,e]

[a] Significantly different from "without" S-LSR; [b] Significantly different from the correspondent maximal velocity; [c] Significantly different from breaststroke ($p<0.05$); [d] Significantly different from butterfly ($p<0.05$); [e] Significantly different from crawl ($p<0.05$)

DISCUSSION: The measured wave heights were partially coherent with previously available results (Ohmichi et al., 1983). Absolute values are of the same magnitude, but the relative butterfly values were higher than expected. Moreover the wave height values only grew with velocity for the front crawl. Both for backstroke and butterfly lower values were found at maximal compared to sub-maximal velocities, suggesting that these ones might be too low to allow proper hydrodynamics. Considering dynamometrical data, butterfly also showed a tendency to produce more intense waves at lower velocities, probably for the same reason. Out of these exceptions, data obtained at lower velocity normally present also lower values. Results also suggest that, at the same velocities, front crawl technique seems to produce less intense waves than the other techniques, including backstroke. Moreover, the generality of the measured wave variables were reduced when the S-LSR was included, showing that these devices are able to significantly attenuate water surface waves during swimming events.

CONCLUSION: This study showed that swimming-line separation ropes are able to significantly attenuate water surface waves during swimming events. Moreover it emphasises that front crawl seems to be the lower wave producer technique among the four swimming competitive strokes at comparable velocities, and that at least butterfly needs to be swam at velocities higher than a critical value to ensure proper hydrodynamics of the swimmer.

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LEVELS OF MUSCLE ACTIVATION IN STRENGTH AND CONDITIONING EXERCISES AND DYNAMOMETER HIKING IN JUNIOR SAILORS

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This study recruited 29 (17 male and 12 female) high-level Byte class sailors aged 14-16 years to examine the average levels of muscle activation in lower limb and trunk muscles in four selected strength and conditioning exercises (leg extension, back squat and back extension exercises, a 30-second hiking hold) and a maximal three-minute hiking test (HM_{180}). Results revealed that between-phase differences existed in the exercises examined. The level of muscle activation for the vastus lateralis for the leg extension exercise was shown to be comparable to that recorded during the back squat. Further, these exercises produced greater amounts of muscle activation when compared to those recorded during the HM_{180} test. Finally, the hiking hold produced greater levels of rectus abdominus activation when compared to the HM_{180} test.

KEYWORDS: sailing, strength and conditioning, EMG

INTRODUCTION: Olympic sailing is typically of an hour's duration and involves sailing 7-11 legs around marker buoys. Due to the unique physical demands of the sport, Olympic sailors are known to have well-developed levels of strength and strength endurance (Larsson et al., 1996). This may also be the case for the "Byte" class which has been selected as the single-handed class for the 2010 Youth Olympic Games. One of the reasons that strength and strength-endurance is well developed in Olympic sailors is due to a manoeuvre called "hiking". Hiking is considered as the most demanding aspect of Olympic class sailing (Larsson et al., 1996). Two hiking positions are typically adopted in Olympic sailing and these include; the short hiking position, where the trunk is kept rigid whilst the knees and hips are flexed and the long hiking position, where the trunk, hips and knees are relatively extended. These postures can be performed either statically or dynamically. The purpose of the sailor adopting a hiking position is to keep the sailing dinghy upright. This is done by counterbalancing the forces generated by the wind on the sail (termed the heeling force) through developing a "righting moment". Important muscles/muscle groups in hiking are thought to include: the medial quadriceps, hamstrings, paraspinal muscles and the abdominals (eg. Larsson et al., 1996; Tan et al., 2006).

It is of importance for sailors to keep themselves injury free whilst maximizing performance. This is especially important for juniors who form the competitive base of the sport and who have the potential to be the next generation of elite athletes. Therefore, more should be understood about the training practices in sailors. For instance, further information could be made available on the demands of hiking in relation to the strength and conditioning exercises that are commonly prescribed in this sport. Collecting biomechanical data in an aquatic and windswept environment is challenging, therefore collection of sport-specific data in a controlled laboratory environment can be justified. To this end, hiking performance has been assessed using a customised hiking dynamometer (Tan et al., 2006). These authors found that maximal hiking performance measured over a three-minute period (the so-called HM_{180} test), was associated with better results in a race.

The purpose of this study was to examine the levels of muscle activation in lower limb and trunk muscles in four selected strength and conditioning exercises and the HM_{180} test. This study was undertaken in 14-16 year old high-level Byte class sailors from Singapore.

METHODS: A total of 29 high-level Byte class sailors aged 14-16 years were recruited from the Singapore National Byte Class Training Squad (n=12, 8 males, 4 females) and the Singapore Byte Class High Participation Group (n=17, 9 males, 8 females). Males were of age 14.1 ± 0.7 years, height 167.8 ± 4.5 cms and mass 55.5 ± 7.7 kg and females were of age 14.3 ± 1.0 years, height 158.6 ± 6.8 cms and mass 51.1 ± 10.0 kg. Ethical approval was obtained from the relevant Institutional Human Research Ethics Committees to conduct the study.

This study involved two testing sessions separated by at least 72 hours. In the first session, participants underwent a six repetition maximum (6RM) strength test for two exercises (back squats and leg extension). The purpose of this session was to set the exercise intensity for these exercises in the second session. Participants from the National Byte Class Training Squad had a minimum of six months resistance training experience, whilst the Byte Class High Participation Group had no structured resistance training experience prior to participation in this study. However, prior to strength testing, the latter group were provided with sufficient familiarisation to both these exercises. Average (± SD) 6RM values were 59.1 ± 17.3 kg and 40.8 ± 13.1 kg for leg extension, and 47.5 ± 15.7 kg and 32.3 ± 12.6 kg for the back squat for males and females participants respectively.

The second testing session involved collecting electromyography (EMG) signals from selected lower limb and trunk muscles whilst participants performed four strength and conditioning exercises used to train junior sailors (leg extension, back squat, back extension, 30 second isometric hiking hold-long hiking position). The mass lifted during the leg extension and back squat exercises was the value recorded for the 6RM strength test. Three sets of each exercise were performed with three repetitions (except the hiking hold as it is a long duration isometric exercise). All relevant exercises and were carried out with a 2-1-2 tempo. These sets were completed in a randomized order to prevent any ordering effect and a three minute rest period was provided between sets to minimise the effect of fatigue. Participants then performed the HM_{180} test (Tan et al., 2006). In this test, participants were requested to hike maximally for the duration of the test on a hiking bench customised to the Byte class. Participants were allowed to adopt long or short hiking postures, jerk, crouch, or alternate their body weight on either leg however, during 30-second analysis periods (see below) long hiking postures were adopted (see below for further details). Whilst performing the abovementioned tasks, EMG signals were collected bilaterally from four muscles (rectus abdominus, superficial lumbar multifidus, vastus lateralis, and biceps femoris) using a portable ME3000 P8 data logger (Mega Electronics®, Kuopio, Finland) operating at 1000 Hz. To identify eccentric and concentric phases (where necessary) during data collection, participants were also filmed using a standard video camera. From this footage and the use of a triggered LED, the timing of the eccentric and concentric phases was determined. For the purpose of EMG data normalisation, participants also performed a series of maximum voluntary isometric contractions (MVICs). Participants performed three, five-second efforts for rectus abdominus and superficial lumbar mulifidus (Dankaerts et al., 2004), vastus lateralis (Lin et al., 2008) and biceps femoris (Mohr et al., 1998).

The muscles of interest for EMG analysis included; back extension (biceps femoris, lumbar multifidus), back squat (bicep femoris lumbar multifidus, vastus lateralis), leg extension (vastus lateralis), isometric hold (rectus abdominus) and the HM_{180} test (rectus abdominus and vastus lateralis). To quantify the level of muscle activation, raw EMG data were demeaned, full-wave rectified and low pass filtered at 4 Hz using a second order Butterworth filter to produce a linear envelope. The MVIC value for each muscle was considered as the greatest mean value recorded for a 200 msec window of the linear envelope measured in any of the three MVIC trials for each muscle. EMG data for the concentric and eccentric phases were then time normalized (0-100%) using cubic spline interpolation and the ensemble average of the three repetitions was calculated. The mean level of muscle activation was then calculated for each muscle of interest for each exercise. For the hiking hold, the mean level of muscle activation was calculated between the 10-15 second period for the hiking hold whilst for the HM_{180} test, the mean level of muscle activation was

calculated between 30-60 seconds, 90-120 seconds and 150-180 second periods. Analyses were conducted with customised software.
To determine between-set reliability for the level of muscle activation, intra-class correlation coefficients (ICC's) were calculated. Three-way ANOVA's with repeated measures with two between-group variables (muscle side, gender) and one repeated measures variable (exercise – which included when appropriate; the strength and conditioning exercises, the concentric/eccentric phases of these exercises and the three periods of the HM_{180} test) were performed for each muscle. Post-hoc analysis with particular emphasis on between-phase differences and between-exercise differences were performed using Least Significant Differences approach. Statistical analysis was performed using SPSS V17.0 for Windows (SPSS Inc, Seattle, WA, USA) with the alpha level set at 0.05.

RESULTS AND DISCUSSION: As all variables showed excellent reliability (ICC values>0.750), data from three sets were averaged for subsequent analysis. Levels of muscle activation for each muscle for the four strength and conditioning exercises and the three HM_{180} test periods are presented in Tables 1 and 2 respectively. Data in these tables have been pooled as there were no between-side or between-gender differences. Significant effects were found for all repeated measures conditions for each muscle (p<0.001).

Table 1. Mean (SD) level of muscle of activation (%MVIC) for muscles of interest. Data are presented for eccentric (ECC) and concentric (CON) phases where appropriate. The hiking hold exercise was isometric (ISO) in nature. Relevant post-hoc analysis results are included.

	Leg Extension		**Back Squat**		**Back Extension**		**Hiking Hold**
	ECC	CON	ECC	CON	ECC	CON	ISO
Biceps Femoris.			31.0* (11.2)	15.2 (5.7)	21.7* (8.4)	34.1 (10.8)	
Lumbar Multifidus			37.7* (11.2)	51.4 (14.6)	24.6* (7.8)	46.1 (16.0)	
Vastus Lateralis	38.9* (12.4)	58.8 (19.5)	37.7* (11.5)	54.0 (17.0)			16.7** (7.1)
Rectus Abdominus							45.3 (22.5)

*Indicates the ECC phase of the exercise was significantly different (p<0.05) to the CON phase.
**Indicates significantly different (p<0.05) when compared to all conditions for this muscle in this table.

Table 2. Mean (SD) level of muscle of activation (%MVIC) for muscles of interest during the three, 30-second periods during the HM_{180} test.

	30-60 sec	**90-120 sec**	**150-180 sec**
Lumbar Multifidus	7.2^,+ (5.3)	7.1^,+ (5.8)	15.6^ (13.0)
Vastus Lateralis	26.5^ (10.3)	22.8^,+,# (9.1)	26.4^ (10.0)
Rectus Abdominus	32.4^ (26.8)	25.2^,# (19.3)	29.3^ (19.5)

^Indicates significantly different (p<0.05) when compared to all conditions in Table 1 for this muscle.
+Indicates significantly different (p<0.05) when compared to the 150-180 sec condition.
# Indicates significantly different (p<0.05) when compared to the 30-60 sec condition.

With an increasing focus being placed on evidence-based practice in exercise and sport science, the quantification of muscle activation in training activities (such as strength and conditioning exercises) and target skills (such as the HM_{180}) test is warranted. In this study there were a number of findings that have practical application.

Firstly, there were numerous significant differences evident between the three periods analysed during the HM_{180} test and the strength and conditioning exercises. These differences in the level of muscle activation clearly showed that the strength and conditioning exercises examined in this study clearly overload the HM_{180} test. There were also several significant findings between the concentric and eccentric phases of each strength and conditioning exercise which was to be expected.
Secondly, whilst not evaluated via statistical analysis, one surprising finding of the study was that the superficial lumbar multifidus showed comparable levels of activation to those reported for the vastus lateralis in the back squat. Also, there were comparable levels of muscle activation in the concentric phase of the back squat and the back extension exercises for this muscle. This may suggest a possible technique problem. The back squat is an exercise that is used to primarily strengthen the quadriceps and the gluteals. Whilst the angle of the trunk was not determined in this study, the superficial lumbar multifidus may have been required to activate more than necessary as the trunk may have displayed excessive flexion in the back squat. In junior athletes, whilst performance can be improved through resistance training, this should not be at the expense of exercise technique. Faulty squatting technique has the potential to expose the athlete's passive structures of the lumbar spine to excessive load.
Thirdly, the vastus lateralis showed similar levels of muscle activation in the leg extension exercise as the back squat, and this may indicate that leg extensions can be used to strengthen the quadriceps whilst squatting technique is developed.
Limitations of this study included; hiking performance was measured on a hiking bench rather than on-water therefore, this study has reduced ecological validity. Further, muscle activation is dependent upon the weight lifted therefore, generalisation of activation levels to lower weight - higher repetition work have different repercussions.

CONCLUSIONS: Leg extension may be an appropriate exercise to increase quadriceps strength in the early phase of strength development whilst squatting technique is refined. Both the leg extension and back squat are capable of providing an overload stimulus for the HM_{180} test. Likewise, the hiking hold can overload the rectus abdominus in the HM_{180} test.

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Sports Injury

EXCESSIVE TIBIAL ROTATION IS RESTORED AFTER ANATOMICAL DOUBLE BUNDLE ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

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Recent studies suggested that tibial rotaion was not restored after single bundle anterior cruciate ligament (ACL) reconstruction. This study aimed to determine if excessive tibial rotation would be restored by anatomical double bundle ACL reconstruction. Ten male subjects with unilateral ACL injury were prospectively assessed with a high demanding task before and after ACL reconstruction. Knee kinematics during pivoting movement was measured by a motion analysis system. The tibial rotation was reduced in the reconstructed knee after ACL reconstruction than the deficient knee ($p<0.05$). There was no difference on tibial rotation between intact knee and reconstructed knee. We concluded that the excessive tibial rotation in the deficient knee was restored after anatomical double bundle ACL reconstruction.

KEYWORDS: Sports medicine, ACL, rotational laxity, functional stability.

INTRODUCTION: Anterior cruciate ligament (ACL) injury leads to knee instability, mainly anterior-posterior (AP) and rotational direction. Clinically, the rotational laxity before and after ACL reconstruction is only examined with subjectively pivot shift test performed by orthopaedic surgeons, which is a combined movement of valgus and external rotation. The objective biomechanical way to evaluate functional rotational stability during high demanding movement is still not widely adopted in clinical setting.
Several kinematics studies, which intact knee was used as a control, investigated patients with unilateral ACL injury. On treadmill running, tibial rotation increased with speed in both injured and normal knees (Czerniecki et al., 1988). Waite and coworkers (2005) reported that low demand activity such as walking and running did not produce sufficient stress to initiate knee instability in ACL deficient knee. In a study to evaluate functional stability after ACL reconstruction, tibial rotation was found not to be restored after single bundle ACL reconstruction with hamstring graft during a pivoting movement (Georgoulis et al., 2007).
Optical motion analysis with reflective skin markers provided common and practical techniques for assessing kinematics of lower limb extremity. It was reported as a widely accepted, well-established and reliable gait analysis method (Chambers & Sutherland, 2002). The purpose of the current study was to investigate the range of tibial rotation of ACL deficient and reconstructed knees during a high demanding task. The contralateral intact knee was used as a control. We hypothesized that there would be a significant excessive tibial rotation in ACL deficient knee and this would be restored by anatomical double bundle ACL reconstruction.

METHOD: Ten male subjects (age = 27.2 ± 4.7yr, height = 1.76 ± 1.00m, body mass = 69.1 ± 9.2kg) with unilateral ACL injury were recruited in the study. ACL rupture was confirmed either by arthroscopy, magnetic resonance imaging or clinical examination. All subjects reported instability during sports and were suggested to have surgical treatment. All injuries were sport-related and all subjects participated at least one time per week of their sports before the injury. The university ethics committee approved the study. Informed consents were obtained from each subject before the study.

All subjects were assessed before and after anatomical double bundle ACL reconstruction with mean follow-up of 329 ± 187 days. An optical motion analysis system with eight cameras (VICON, UK) was used to record three dimensional movements of lower extremities at 120Hz frequency. Synchronized force-plate (AMTI, USA) data was collected at the centre of the capture volume at 1080Hz. A fifteen-marker model (Davis et al., 1991) was adopted to collect lower limb kinematics during the movement. Skin reflective markers with 9mm diameter were placed at anatomical landmarks including anterior superior iliac spines, sacrum, greater trochanter, femoral epicondyle, tibial tubercle, lateral malleolus, heel and fifth metatarsal head on both limbs.

Before performing the movement, a standing trial was recorded for each subject in anatomical position. This calibration file provided a definition of zero degree for all segmental movements. Both deficient and intact knees were tested individually. The subjects were asked to jump from a 40cm-high platform and land with both feet on the ground, with only the testing foot on the force-plate. After the foot contact, the subjects pivoted 90 degrees to the lateral side of testing leg, which also acted as the core leg during pivoting. The subjects were instructed to run away after completing the pivoting movement. The evaluation period was defined from the first foot contact to the take-off of the testing leg on the ground. Three dimensional coordinates of every marker were exported from the VICON software. Together with the anthropometric measurements, the joint kinematics was then calculated (Davis et al., 1991). All calculations were conducted using self compiled program (MATLAB, USA). Force-plate was used to determine the evaluation period when the ground reaction force exceeded 5% of body mass. The main dependent variable in the current study was the range of tibial rotation angle during pivoting movement, which was a period defined from the lowest tibial internal rotation after landing to the highest tibial internal rotation (Ristanis et al., 2005).

A paired t-test was employed to determine if statistically significant difference existed in range of tibial rotation between intact knee and deficient knee before reconstruction, between deficient knee and reconstructed knee, and between reconstructed knee and intact knee after reconstruction. Power analysis was also conducted if there was no significant difference between reconstructed knee and intact knee after reconstruction. The level of significance and study power were set at 0.05 and 0.8 respectively.

RESULTS: During the pivoting movement, the tibia internally rotated to a maximum degree (Figure 1). For the range of tibial rotation, there was a significant increase in the deficient knee (12.6 ± 4.5 degree) when compared to the intact knee (7.9 ± 3.1 degree) before reconstruction. This excessive tibial rotation significantly decreased to 8.9 ± 3.0 degree for the reconstructed knee and did not differ to that of intact knee (8.2 ± 2.6 degree) after ACL reconstruction (Figure 2). Since there was no significant difference between reconstructed knee and intact knee after reconstruction, power analysis was conducted (true difference: 2 degrees; correlation: 0.27) and statistical power was reported to be 0.81 between the two groups.

DISCUSSION: In this study, the excessive tibial movement for ACL deficient knee and the reduction of this movement after ACL reconstruction were demonstrated. It was hypothesized that there would be a significant excessive tibial rotation on ACL deficient knee and it would be restored by anatomical double bundle ACL reconstruction. The result of the current study supported the first hypothesis. Moreover, the study power between both limbs after ACL reconstruction was above the pre-set value and so the second hypothesis was also supported in this study.

Our findings supported previous studies (Zaffagnini et al., 2000; Ristanis et al., 2005) that showed knee rotational laxity and instability of ACL deficient knee. In a swine study (Zaffagnini et al., 2000), the passive clinical internal-external knee rotation stress test was shown to give excessive laxity of ACL deficient knee when compared to intact knee. In another study (Ristanis et al., 2005) with similar protocol to the present study, the tibial

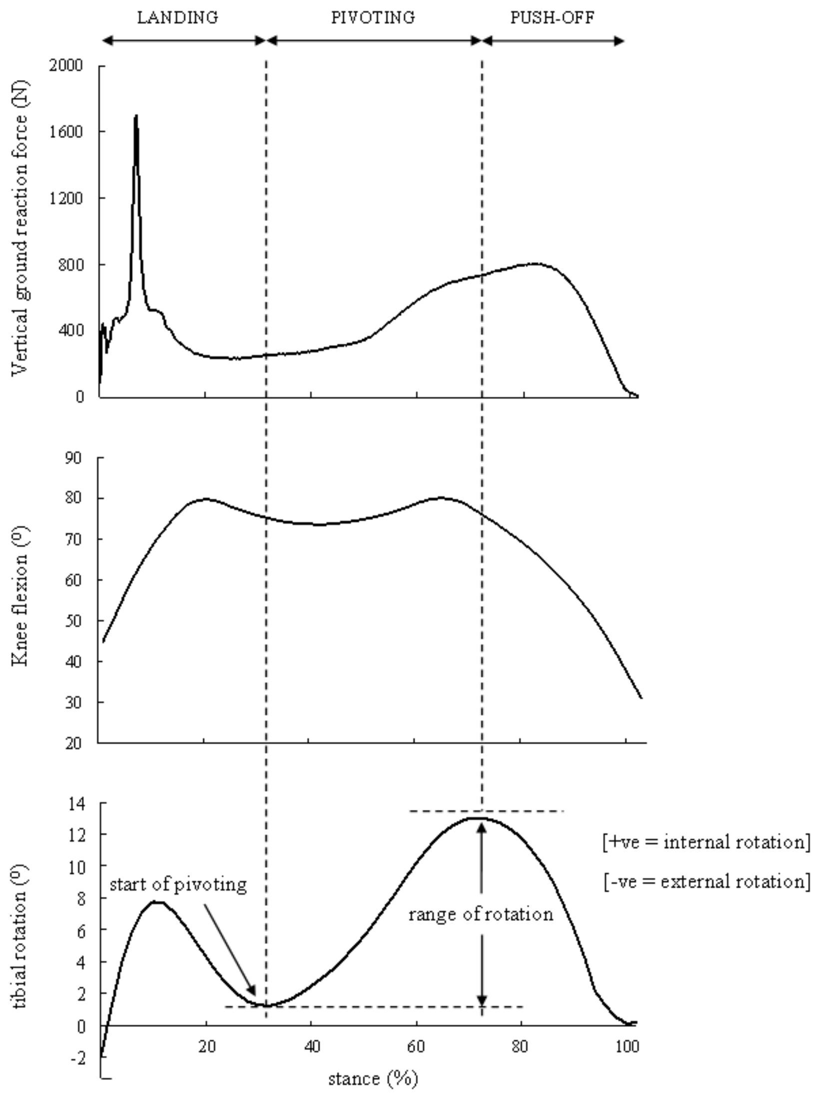

Figure 1. Ground reaction force, flexion angle and tibial rotation during the entire stance phase of the high demanding task.

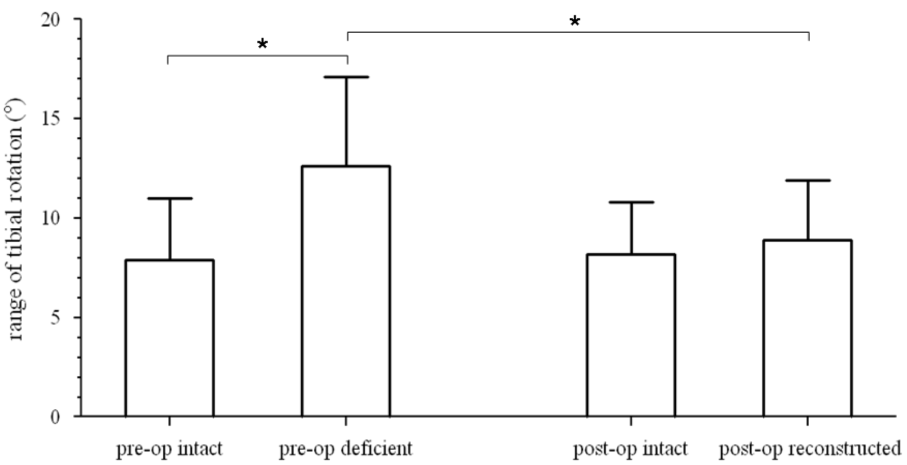

Figure 2. Range of tibial rotation during pivoting movement before and after ACL reconstruction. Asterisks (*) indicate a significant difference.

rotation of deficient knee was significantly higher than that of intact knee. While those subjects were instructed to walk followed by the pivoting movement, our subjects were instructed to run instead. We believed that the task in our study provided a higher rotational stress to the knee. Moreover, in our study, all the subjects were assessed prospectively before and after ACL reconstruction. The variations between study and control groups were limited to affect our result as intact knee was used as a control.
ACL reconstruction aims to reconstruct the original ACL with normal kinematics in both AP and rotational direction. However, in vitro (Woo et al., 2002) and in vivo (Georgoulis et al., 2007) studies showed that tibial rotation was not restored by single bundle ACL reconstruction. One of the reasons suggested that only anteromedial (AM) bundle was replicated, resulting in insufficient rotational control to the knee. In the current study, all subjects were treated with anatomical double bundle ACL reconstruction, in which both AM and posterolateral (PL) bundles were reconstructed to mimic the original ACL anatomy. In addition to the AM bundle, PL bundle provided an important role in the stabilization of the knee against a combined rotatory load (Gabriel et al., 2004). When evaluating double bundle ACL reconstruction with a high demanding activity, the significant decrease in range of tibial rotation of the reconstructed knee suggested the effectiveness of rotational control of PL bundle.

CONCLUSION: It was concluded that there was excessive tibial rotation in ACL deficient knee during a dynamic functional pivoting movement in this study. The reconstructed knee successfully improved functional knee rotational stability as demonstrated by the restoration of excessive tibial rotation during a pivoting movement before and after anatomical double bundle ACL reconstruction.

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PREDICTION OF ANKLE JOINT TORQUES USING ARTIFICIAL NEURAL NETWORKS

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Major ankle sprains in sports are thought to be due to high levels of ankle torsion. The purpose of this study was to develop a method for measuring *in vivo* ankle torques developed by athletes. Motion capture, force plate, and insole pressure measurements were used to develop generalized regression neural networks to predict maximum ankle torque and rate of ankle torque based on insole pressures. It was found that network prediction accuracy depended on the number of subjects used for training, as well as the method of pressure sensor grouping. Further work will be performed to determine optimal subject and pressure sensor groupings.

KEY WORDS: ankle torque, sports injury, neural network, motion capture, insole pressure

INTRODUCTION: Ankle injuries are prevalent in sports. Among these injuries are a group referred to as ‘eversion’ ankle sprains or ‘high ankle sprains’. While these injuries are less frequent than the ‘inversion’ sprain, they require a longer time for rehabilitation (Williams et al., 2007). Clinical (Boytim et al., 1991) and experimental (Villwock et al., 2009) studies suggest high ankle sprains occur due to excessive torque, or external rotation of the foot. Recent studies by Villwock et al suggest that the shoe-surface interface conditions may be important factors in the generation of high levels of external torque on the ankle. However, the levels of torque developed by athletes on playing surfaces are unknown at this time. The objective of this study was to investigate the use of an in-shoe pressure measurement system to determine the maximum torque as well as rate of torque generation developed by subjects twisting under controlled laboratory conditions. It is believed that maximum torque can be a predictor of injury while the rate of torque generation is directly related to performance. In sports it is important to optimize performance while minimizing the risk of injury. The findings of the study will assist in the development of techniques to determine torques developed by athletes with various shoe designs on various types of synthetic and natural turfs.

METHODS: Data from two male subjects, performing ten trials each, were used in the current study. A six-camera Vicon MX Motion Capture System (OMG plc., Oxford, UK) was used to capture 3-dimensional marker data at a sample rate of 100 Hz. The standard lower body Vicon Plug-in Gait marker set was used for this study. A force plate sampling at 1000 Hz (Advanced Medical Technology, Inc., Watertown, MA) was used to measure ground reaction forces. Plantar pressures were measured with 24 sensors in the Parotec insole pressure measurement system (Paromed, Neubern, Germany). Prior to any recorded trials, the subject was instructed on how to perform the torque motion. Subjects stood with their feet shoulder width apart with the right foot on the force plate and their knees slightly bent. The subject then internally rotated his right shank with respect to the stationary right foot. Internal/external ankle moment data was recorded using the Plug-in Gait kinetic and kinematic packages. The right Parotec insole data and the output ankle torque values were put into a file for each subject trial. The method used to model the maximum torque and rate of torque generation was a neural network.
Generalized regression neural networks (GRNN) have the ability to model complex functional mappings between inputs and outputs. However, GRNN’s suffer from dimensionality which is characterized as a possible overabundance of input variables (Kiyan,

2004). In order to investigate optimization of dimensionality of the input vector from 24 individual sensor inputs, the pressure sensors were grouped four different ways based on their relative contribution to the output torque. All sensor groupings took into account knowledge that the pressure distribution during the torque motion changes from uniform pressure distribution across the foot to more pressure on the medial side. Using this premise, sensor grouping A divided the insole into medial and lateral halves, leaving out the sensors that fell along the center line (sensors 8 and 11). Sensor grouping B divided the sensor data into three groups, medial, lateral, and a center group. It was assumed that this grouping would improve upon Sensor Grouping A by adding a group in the center to track the lateral to medial pressure change. Groupings C and D further refined the clusters by breaking up sensor groupings across the foot. This was done to allow the networks to have better prediction capabilities due to possible changes in pressure in the anterior/posterior direction during the motion.

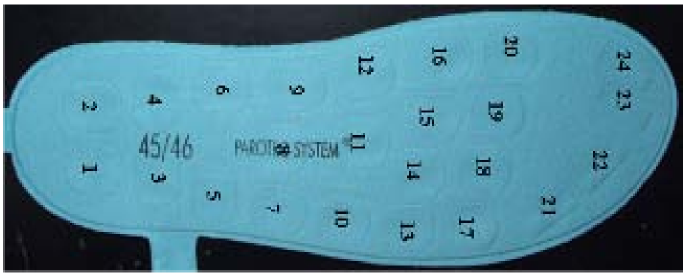

Figure 1: A Numbered Parotec Insole (right foot)

Table 1: Four different sensor groupings

Sensor Grouping A (2 Dimensions)	**Sensor Grouping C (8 Dimensions)**		
1,3,5,7,10,13,14,17,18,21,22	1,3	2,4	5,7,10
2,4,6,9,12,15,16,19,20,23,24	6,9,12	21,22	23,24
	13,14,17,18		15,16,19,20
Sensor Grouping B (3 Dimensions)	**Sensor Grouping D (11 Dimensions)**		
1,3,5,7,10,13,17,21,22	1,3	2,4	5,7,10
8,11,14,15,18,19	6,9,12	8,11	13,17
2,4,6,9,12,16,20,23,24	14,18	15,19	16,20
	21,22		23,24

In order to develop a neural network, the network must first be trained using representative data. Four different networks were trained and tested to determine which led to the most accurate prediction of maximum torque and rate of torque generation. The first two networks used data from both subjects for training and one subject to test, i.e. 19 of the 20 trials were used to train and 1 trial was chosen at random to test the network’s ability to predict the torque data based on the insole pressure measurements alone. The other two networks were trained and tested with each subject’s data independently, i.e. 9 of the 10 trials from one subject were used to train and 1 out of 10 trials was used to test. Each of these four networks was run with four different sensor groupings. A total of sixteen neural networks were constructed and tested. The success of each network was determined by how accurately the actual and predicted curves aligned, based on an error analysis. The time trace error value represented the square root sum of the squares between the predicted and actual torque values at individual time points. The peak torque error represented the difference between the actual and predicted maximum torque values.

RESULTS: The time trace error values, as well as the peak torque errors, are shown in Table 2 for each of the networks. Table 3 provides averages for the error values produced by each sensor grouping. The smallest error values were seen when both subjects were used to train the network, Networks 1 and 2. Sensor grouping D produced the lowest time trace error value regardless of the network. Looking specifically at Network 1, sensor grouping D was best at predicting the rate of loading with an error value of 15.00. Sensor grouping A was best at predicting the peak torque, its average error value across all networks was 1.71%. Looking specifically at Network 2, the difference between the actual and predicted maximum torque values for sensor grouping A was 0.07%.

Table 2: Error values recorded for each network and sensor grouping

	Subject(s) used for Training	Subject used for testing	Sensor Grouping	Time Trace Error (Nm)	Peak Torque Error (%)
Network 1	1,2	1	A	19.23	2.70
	1,2	1	B	16.72	14.28
	1,2	1	C	18.68	14.75
	1,2	1	D	15.00	11.39
Network 2	1,2	2	A	34.49	0.07
	1,2	2	B	18.22	4.67
	1,2	2	C	18.65	2.29
	1,2	2	D	17.94	3.35
Network 3	1	1	A	27.52	2.73
	1	1	B	20.94	13.78
	1	1	C	28.45	2.80
	1	1	D	17.61	0.73
Network 4	2	2	A	18.22	1.37
	2	2	B	19.52	0.45
	2	2	C	18.66	2.29
	2	2	D	17.94	3.35

Table 3: Average Error Values for sensor groupings

Sensor Grouping	Time Trace Error (Nm)	Peak Torque Error (%)
A	24.87	1.72
B	18.85	8.30
C	21.11	5.53
D	17.12	4.71

DISCUSSION: The findings support the notion that a properly trained neural network could be used successfully to predict torque values based on insole pressure measurements alone. It was found that a network trained on two subjects was more precise than those that were trained and tested on only one subject. This is supported in other studies that have shown networks tend to be most accurate predictors when they are trained on data from many test subjects (Haykin, 1999). The current study also showed that the prediction accuracy of maximum torque and rate of torque generation was greatly affected by sensor groupings. Grouping A, which had two dimensions (medial-lateral), best predicted the maximum torque (Figure 2b). This may be due to the fact that at peak torque, the pressure distribution was concentrated on the medial border of the insole. Sensor grouping D, which was based on eleven dimensions, best predicted the rate of loading (Figure 2a).

** Actual values calculated from inverse dynamics, predicted values calculated from the neural network

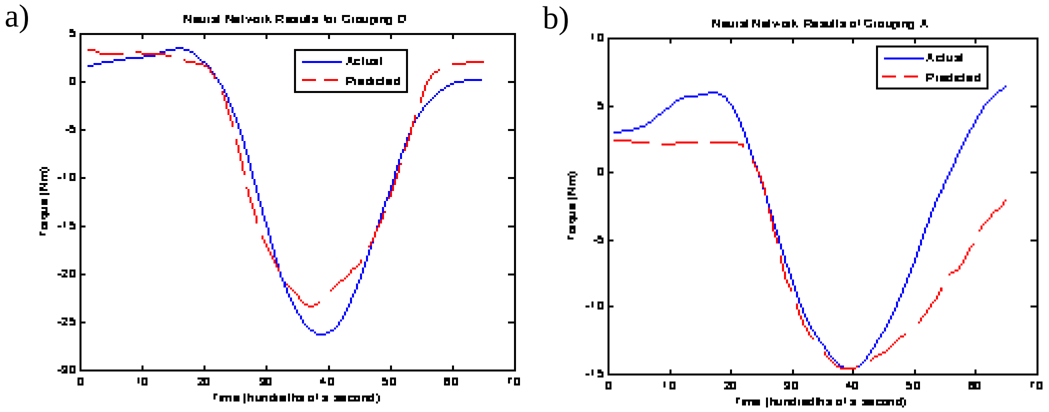

Figure 2: Torque results from training with Subjects 1 and 2, a) and tests on Subject 1 in sensor grouping D, b) and tests on Subject 2 in sensor grouping A

This result may suggest that adding dimensions may be required to better predict temporal torque data. (Erkmen & Yildirim, 2008). It is not uncommon to use different prediction methods and levels of dimensions to optimize the prediction capabilities of various physical parameters, based on pressure insole data (Cordero et al, 2004).

CONCLUSION: This study demonstrates that a neural network based on insole pressures could successfully predict ankle torque and rate of torque generation. Findings indicate that different sensor groupings influenced the accuracy of the neural network. Depending on the aim of the tests, an optimal sensor grouping could be selected. Future work should be continued to determine the optimal sensor grouping for both maximum torque and rate of torque loading, as well as determine if an individualized network will produce more accurate outputs than combined subject networks. The use of artificial neural networks to predict torque and rate of torque generation by subjects on current playing surfaces with various shoes could prove valuable in determining optimal combinations of shoes and playing surfaces. Future studies may utilize this testing technology to develop combinations of shoes and playing surfaces that optimize performance and limit injury risk.

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LUMBAR KINEMATICS AND KINETICS OF YOUNG AUSTRALIAN FAST BOWLERS

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Lower back injuries are a serious concern for cricket fast bowlers. As lumbar loading is the causal mechanism of such injuries, the purpose of this study was to find relationships between lumbar loads and selected kinematic variables. Thirteen young fast bowlers (17.4 ± 1.9 years) were tested with a 3-D motion analysis system (200 Hz). Kinematics and lumbar spine kinetics were calculated about the L5/S1 joint during the arm acceleration phase. The largest kinetic values were the lumbar axial forces and lumbar flexion moments. Maximum lumbar spine moments were associated with several kinematic variables such as front knee angle, pelvic and thoracic rotation at ball release, and shoulder counter-rotation. Modifying bowling kinematics may reduce lumbar loads and reduce the potential for lower back injuries.

KEYWORDS: cricket bowling, lumbar spine, counter-rotation, lumbar kinetics, fast bowling.

INTRODUCTION: Fast bowling in cricket requires the bowler to rapidly flex, laterally bend and rotate the lumbar spine in order to produce ball speeds up to 45 m s^{-1}. These movement patterns are considered to play a role in the development of lower back injuries. Intervertebral disc abnormalities and soft tissue injuries in the lumbar region are often observed in fast bowling populations, but the most serious condition in terms of lost playing time involves fractures to the pars interarticulares, particularly of the L4/L5 or L5/S1 vertebrae (Elliott et al., 1993; Portus et al., 2004; Orchard, 2006). The problem is of great concern to cricket administrators and coaches, because young bowlers are the most at risk group (Portus et al., 2007).
Researchers have studied various kinematic factors for associations with lumbar injury incidence. A number of studies have found there to be an increased risk of lumbar spine injury when shoulder counter-rotation, a preliminary rotation of the shoulder girdle in the horizontal plane away from the direction of bowling, is in excess of 30° or 40° (Elliott, 2000; Portus et al., 2004). Pelvic-shoulder separation angle at back foot angle has been associated with a moderate increase in soft tissue injury (Portus et al., 2004). In addition, bowlers with back injuries may utilise greater ranges of lateral flexion of the lumbar spine during delivery stride (Portus et al., 2007). In terms of identifying the causal mechanisms of lumbar injury, kinetic calculations are required. Ferdinands et al. (2009) tested 21 fast bowlers of premier grade level and above and found that large flexion, rotation and lateral bending moments were placed on the spine when displaced towards the end of its available range of motion. However, there has been no study to date that has investigated the association between kinematic variables and lumbar spine loads.
The established method of determining lumbar injury risk in fast bowlers is mostly based on shoulder counter-rotation, which is only a kinematic measure. Hence, shoulder counter-rotation is not a causal mechanism of lumbar injury. By establishing the kinematic correlates of lumbar spine loads, it may be possible to develop a more accurate assessment of lumbar injury risk. The identification of these kinematic characteristics may have implications for the development of safer bowling techniques, particularly with respect to younger bowlers. Therefore, the purpose of this study was to use three-dimensional motion analysis and inverse dynamics to investigate the relationships between kinematics and lumbar spine kinetics within an elite sample of young fast bowlers. The hypothesis is that there are associations between lumbar loads and kinematic variables, particularly with those kinematic variables that have been associated with an increased incidence of lumbar injury.

METHOD: Thirteen young fast bowlers (17.4 ± 1.9 years) were recruited from the Cricket New South Wales development squad. The trials were performed in a biomechanics laboratory, which permitted a full length run-up. A 14-camera Cortex Motion Analysis System (Version 1.0, Motion Analysis Corporation Ltd., USA) was used to capture three-dimensional (3D) motion (200 Hz) and force plate (1000 Hz) data on 20 trials for each bowler while front and rear foot contact was made on two Kistler force plates. Each subject was instructed to bowl at maximum effort as in match conditions. Five trials in which the ball landed within a 'good length' area demarcated by two white lines 13 m and 19 m from the stumps at the bowler's end were selected for analysis. Subjects also rated their performance from 0 to 10 using an analogue performance scale. The video capture volume encompassed the back foot contact, front foot contact, ball release and follow through phases of the bowling action. The Cortex system was calibrated according to the manufacturer's recommendations resulting in a residual error of marker position of less than 1 mm.

Motion analysis capture was performed on each subject wearing a full body marker set comprising forty-five 15 mm spherical markers, which were attached to bony landmarks (Ferdinands, 2009). Markers were located on the left and right sides of the body except for markers half-way between the posterior superior iliac spines (mid-PSIS), and on the 7th cervical vertebrae, supra-sternal notch, and the head. The positions of the anterior superior iliac spine (ASIS), mid-PSIS, and greater trochanter markers were used to calculate the hip joint centres. All other joint centres were calculated as the average position between two markers placed either medially and laterally or anteriorly and posteriorly on the joint. Exceptions were the position of the shoulder, mid-trunk, hip markers, and cricket ball. A recursive fourth-order low-pass Butterworth filter was used to smooth the motion analysis data. The cut-off frequencies (8 – 15 Hz) were determined from residual analyses.

The three-dimensional motion analysis data of the markers were imported into a 22 segment rigid body model of the cricket fast bowler in Kintrak (V.7.0, University of Calgary), which is a software programme designed to perform kinematics and inverse dynamics analysis using motion analysis and force plate data. Local segment coordinate systems of the rigid body model were defined for each of their respective segments. The lumbar spine segment (LSS) was defined as a single segment having its inferior end located half-way between the hip joint centres at the level of L5/S1. The superior endpoint was located at the mid-point between the markers on the suprasternal notch and T7.

Kinematic and kinetic data were calculated during the arm acceleration phase defined from the time of maximum vertical front foot ground reaction force to time of maximum hand velocity. Net joint torque was calculated about the inferior end of the lumbar spine segment for lateral bending, flexion/extension and rotation. Right lateral bending, extension and right rotation were defined as positive. Pearson's correlation coefficients were calculated in SPSS (Version 17, SPSS Inc.) to assess the relationship between selected kinematic and kinetic variables.

RESULTS: The sample had the following mean kinematic values: shoulder counter-rotation (39.3 ± 12.7°), pelvic-shoulder separation angle (-23.1 ± 8.2°), thoracic lateral bending (41.5 ± 8.5°), thoracic rotation (119.2 ± 14.6°), pelvic rotation (107.4 ± 13.2°), front knee flexion angle (-17.0 ± 6.5), stride angle (5.0 ± 5.8°) and bowling hand velocity (23.8 ±1.2 m s^{-1}).

Forces and moments were expressed in terms of body weight (BW) and body weight x height (BW m). The highest ground reaction forces were the vertical components (5.3 ± 0.8 and 2.3 ± 0.4 BW) at the front foot and back foot. The mean maximum lumbar forces were 8.0 ± 1.2 BW along the inferior-superior long axis, 1.5 ± 0.6 BW along the posterior-anterior axis and 0.4 ± 0.5 BW along the lateral-medial axis. The mean maximum lumbar torques were 3.1 ± 0.5 BW m (flexion), 0.9 ± 0.4 BW m (left lateral bending) and 0.2 ± 0.2 BW m (right rotation).

Table 1 shows the Pearson correlation coefficients between the lumbar spine kinetic and kinematic variables. There were also other kinematic variables that were not significantly correlated with any kinetics variables: kinematic crunch factor (thoracic lateral bending x pelvic rotation), stride length, and centre of mass horizontal and vertical velocities at back foot contact. Bowling hand velocity was not correlated with any of the above kinematic variables. Counter-rotation was correlated with stride angle ($r = -0.57$, $p = 0.042$). The coefficients of variation of the kinematic and kinetic data were less than 8.0% and 15.2%, respectively.

Table 1. Pearson's correlation coefficients between lumbar spine kinetic and kinetic variables Thoracic and pelvic kinematics were calculated at ball release. Front knee angle was calculated at the time of maximum front foot ground reaction force. (Significance level, p < 0.05)

	Axial Force	Anterior-posterior Force	Medio-lateral Force	Flexion Moment	Rotation Moment	Lateral Bend Moment
Thoracic flexion		r = -0.82 p < 0.001			r = 0.52 p = 0.067	
Thoracic rotation		r = -0.57 p = 0.040		r = 0.59 p = 0.033	r = 0.59 p = 0.034	r = -0.83 p < 0.001
Thoracic lateral bend						r = 0.59 p = 0.035
Pelvic rotation		r = -0.58 p = 0.037		r = 0.49 p = 0.093	r = 0.112 p = 0.056	r = -0.92 p = 0.001
Counter-rotation			r = 0.52 p = 0.067		r = 0.61 p = 0.028	
Pelvic-shoulder separation						r = -0.54 p = 0.053
Front knee angle		r = -0.82 p < 0.001			r = 0.91 p < 0.001	r = -0.81 p = 0.001
Stride angle			r^2 = -0.61 p = 0.028			
Hand velocity			r = 0.50 p = 0.082			
Front foot GRF Z	r = 0.59 p = 0.03					

DISCUSSION: The aim of this study was to investigate the relationships between lumbar spine kinematics and kinetics in a sample of elite young fast bowlers. A relatively young cohort was selected because this age group has a high incidence of lumbar injury (Portus et al., 2007).

Interestingly, the kinematic variables that were most strongly correlated with lumbar spine kinetics were the pelvic and thoracic rotations at ball release. These variables were correlated with four kinetics variables: anterior-posterior force and all three lumbar spine moments. The position of the thorax in bowling and related motions such as throwing all require the pelvis and thorax to face the target at the time of release. However, there is no need for these segments to rotate beyond this as these variables were not correlated with bowling hand speed.

Conversely, the kinematic variables that have previously been associated with an increased incidence of lumbar injury had fewer associations with lumbar kinetics. Shoulder counter-rotation was strongly correlated with the lumbar spine rotation moment, but this moment was very small in magnitude and may not be of clinical significance. In addition, shoulder counter-rotation was strongly correlated with the medio-lateral force, but this may just result from an unbalanced position during delivery stride since counter-rotation was also correlated with a

closed stride angle. Pelvic shoulder separation angle was strongly correlated with the lumbar lateral bending moment. The crunch factor is used as an index of shear stress loads at the lumbar vertebrae. However, this factor was not associated with any lumbar spine moments.
The front leg acts as a shock absorber to attenuate the ground reaction forces upon front foot contact. Knee flexion angle had no effect on the attenuation of the large axial forces, which were strongly correlated with vertical ground reaction force acting through the front foot. However, the front knee flexion angle did have strong correlations with both the lumbar spine rotation and lateral bending moments and therefore has an important effect on lumbar spine loading.
In general, the data shows that there a number of kinematic variables associated with lumbar spinal loads. A combination of these variables can work additively to increase lumbar spine loading. For instance, the data suggests that a bowler landing with a more flexed front knee, large pelvic-shoulder separation angle and large range of lateral thoracic bending has three factors that would contribute to the generation of a lateral bending moment. This suggests that a risk of assessment of lumbar injury in bowling may need to consider multiple variables.

CONCLUSION: This study supports the hypothesis that lumbar loading in fast bowling is associated with kinematic variables. As the causal mechanisms of lumbar injury are ultimately linked to spinal loading, the identification of kinematic variables associated with such spinal loading can lead to an improved assessment of lumbar injury risk in young fast bowlers. However, a prospective longitudinal study is needed to compare a wide range of kinematic and lumbar spine kinetics variables to assess their predictive ability of lumbar spine injury. The researchers are currently evaluating the MRI scans taken of all bowlers towards the end of the cricket season to quantitatively assess their injury status. In addition, the data of another five bowlers will be analysed. Such research has the potential to yield an accurate multi-index assessment of lumbar injury risk. This would give coaches the ability to more accurately screen young bowlers for injury risk and also suggest changes to the kinematics of bowling techniques to reduce lumbar spine loads.

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THE APPLICATION OF A SPORT-SPECIFIC 3D STEREOSCOPIC STIMULUS TO EXAMINE PRE-PLANNING TIME AND GAZE CHARACTERISTICS DURING EVASIVE SIDE-STEPPING MANOEUVRES.

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KEYWORDS: knee loading, 3D stereoscopic stimulus, injury prevention.

INTRODUCTION: It is well established that anterior cruciate ligament (ACL) injuries are serious, debilitating and costly for an individual, while also creating a significant public health burden at a societal level. ACL injuries occur when inappropriate external loads are applied to the knee and most commonly occur during the performance of a side-stepping (Ssg) manoeuvre (Besier et al., 2001a). Previous laboratory based investigations of evasive Ssg have employed generic light or mannequin visual stimuli in an effort to simulate the time and space constraints experienced by athletes, in the preparation and execution of the Ssg manoeuvre (Besier et al., 2001b; Besier et al., 2003; Mclean et al., 2004). However, a possible outcome of attempts to impose these constraints in lab environments is that the use of unrealistic visual stimuli may not accurately reflect or identify the relationship of the perceptual demands of the task with injury risk variables, during a sidestep in game based situations. This study proposes that the presentation of a three dimensional (3D) stereoscopic stimulus (3DSS), featuring a 3D video based sport specific reconstruction of an opposing defender(s) simulating a tackle, may improve the ecological validity of laboratory based investigations. Additionally, the incorporation of the 3DSS tool with eye tracking will allow for the subject's gaze characteristics (fixations, durations on the 3DSS image) to be assessed. The general aims of this study were to:

Technical

- develop a 3DSS that delivers realistic sport-specific constraints to footballers during evasive Ssg manoeuvres,
- create 3DSS scenarios that incorporate realistic game based variations of imposed time and space constraints (e.g. 3DSS tackle scenarios with one or two defenders),
- develop an interface and protocol that integrates the 3DSS system with a commercial eye tracking system (ASL Eye Tracking Recorder), for the purpose of quantifying the lab based subject's gaze characteristics on the projected stimulus during a Ssg manoeuvre,

Experimental

- identify differences in biomechanical variables (kinematic, kinetic and neuromuscular) associated with increased injury risk during Ssg manoeuvres, using a traditional light (light emitting diode) based stimulus (LBSS) compared with a 3DSS stimuli.

METHOD: The 3DSS, featuring scenarios of single and multiple tacklers on a football field, was developed in collaboration with the Western Australian Supercomputer Program (Mr Paul Bourke). While stereoscopic projection is not a new technology, the developed projection format allowed for the scaling of two converging fields of view (FOV). The first FOV was the 3D stereoscopic projection (stimuli) of an oncoming opponent performing a tackle, while the second converging FOV comprised a 'real' lab based footballer performing an evasive Ssg manoeuvre of the 3DSS stimuli.

Two stimuli conditions were examined: 1) the developed 3DDS and 2) an LED based stimuli (LBSS) consisting of a panel of three lights, one of which was illuminated to indicate the required direction of sidestep to be performed. A 12 camera Vicon motion capture system operating at 250 Hz and a 1200 mm x 1200 mm AMTI force plate sampling at 2000 Hz were

used to collect 3D kinematic and kinetic data. A wireless Noraxon EMG system collected surface EMG from selected musculature of the upper and lower body while an ASL Eye Tracking Recorder was calibrated to each subject's field of view and recorded eye gaze fixations and durations on the projected tackler. 3D Kinematic and kinetic data was modelled in Vicon Nexus software using the customised marker set and model of Besier et al, 2003.
Eight healthy, male football players, aged 18-30 years old with no history of major lower limb injury or disease were recruited to take part in the study. Participants performed a planned and unplanned side step using both the light based and stereoscopic stimuli. In the planned scenarios for both stimuli, the participants were told prior to trial commencement the required direction of the Ssg manoeuvre. In the unplanned LBS scenario, the participants were required to react to one of two lights which was illuminated only after the participant triggered the timing gate positioned immediately in front of the force plate. The light was illuminated if the participant's approach velocity met a threshold of $4.5ms^{-1}$ (± 0.2). In the unplanned 3DSS scenario, the projection of the defender/defenders tackling scenario commenced once the timing gate was triggered and the approach velocity threshold range was confirmed. Participants were then required to sidestep to avoid the oncoming 3DSS tackler (e.g. sidestep to the left when the tackler(s) is coming from the right). The presentation order of the stimuli condition (3DSS v LBS and planned v unplanned) was randomised as was the requirement for the subject's to perform sidesteps to the left or right. Three trials were collected for the left and right leg during a Ssg manoeuvre across all four stimuli conditions, comprising 24 trials in total.

RESULTS: Initial results revealed an increase in knee joint moments with trends toward lower levels of co-contraction of the knee stabilising musculature in the unplanned condition for both the LBS and the 3DSS stimuli. Across stimuli conditions, an increase in the number of gaze fixations and accompanying decrease in the duration of fixations was found in the 3DSS stimuli condition. This finding was accompanied by a trend toward slightly earlier muscle activation firing patterns in the 3DSS condition.

CONCLUSION: This study outlines a method for improving the ecological validity of lab based visual stimuli to study ACL injury mechanisms. This was achieved via the presentation of 3D stereoscopic projections of game specific opponents. It appears that the introduction of more realistic stimuli may affects spatial and temporal pre-planning of various neuromuscular biomechanical variables associated with increased knee loading and therefore inferred ACL injury risk. Additionally, the introduction of a more game realistic visual stimulus was shown to affect the athlete's visual search patterns and therefore, by inference, alter their cognitive processing of environmental cues. These findings may be of importance when designing injury prevention training programs.

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AMATEUR BOXER BIOMECHANICS AND PUNCH FORCE

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The current study investigates the correlation between punch biomechanics and punch force in amateur male boxers (n=39). A Hybrid III 50th percentile male dummy was used to gather punch force values. TrackEye Motion Analysis (TEMA) was used to measure the velocity of each boxer's punch. Lower body force values were determined using the Functional Assessment of Biomechanics (FAB) system. Two types of punches, hooks and straights, were analyzed. It was determined that punch forces correlated more strongly to hand velocity than to lower body forces. Punch force correlated to hand velocity with R^2 values of 0.380 and 0.391 for hook and straight punches, respectively ($p<0.001$). Punch force correlated to lower body forces with R^2 values of 0.103 and 0.099, respectively ($p<0.05$).

KEYWORDS: boxing, biomechanics, punch force.

INTRODUCTION: Boxing is a physically and mentally demanding contact sport. Boxers are required to possess a combination of endurance, strength, stamina, agility, coordination, and speed. Amateur boxing is based on a scoring system where the main objective is to impact the opponent while protecting one's self from impacts. The most desirable outcome is to knock out one's opponent, ensuring a win. In the process boxers sustain numerous punches to the head causing minor or even severe injuries. Dedication and intense training is vital for both amateur and professional boxers to establish effective technique.

Previously published studies have focused on boxing related injuries (Porter and O'Brien 1996; Zazryn *et al.* 2006) and the effects of punch biomechanics on the recipient (Sherman *et al.* 2004; Walilko *et al.* 2005). Few studies have evaluated the dynamics that occur throughout the body prior to impacting the opponent (Whiting *et al.* 1988). This knowledge may provide insight relevant to both boxers and trainers regarding effective technique, improving the athlete's skills, and decreasing the number of injuries.

This study evaluates the biomechanics of a punch and its correlation to punch force. Hand velocity and lower body forces were analyzed. To accomplish this goal, male amateur boxers were instructed to impact a Hybrid III (HIII) 50th percentile male dummy. Boxers threw two types of punches—hook and straight. The punch force was calculated using data gathered by the HIII. Pre-impact hand velocity was measured using TrackEye Motion Analysis (TEMA). Functional Assessment of Biomechanics (FAB), a novel motion analysis system, calculated the force values generated by the lower body.

METHODS: A total of 42 adult male amateur boxers participated in this study, which took place at the 2009 Ringside World Championship Boxing Tournament (Kansas City, MO). From the data collected, a total of 39 boxers provided reliable data for analysis. Boxer mass ranged from 54 kg to 118 kg (mean of 77±15 kg). Boxer height ranged from 1.60 m to 1.98 m (mean of 1.77±0.08 m). Prior to recruiting participants, approval was granted from Wayne State University's Human Investigation Committee.

A HIII 50th percentile male dummy (head, neck, and torso) with a frangible face was used in a manner similar to a previous study (Walilko, *et al*. 2005). The frangible face provides boxers with a compliant impact surface that was used to prevent participant injuries. The headform was attached to the dummy neck, which was mounted to the surrogate upper body to ensure a biofidelic headform motion (Viano and Pellman 2005). The upper body was mounted to an adjustable height metal table. Weights were used on the table top and at the bottom of the table legs to prevent the table from lifting off the ground or sliding across it.

Three Endevco (San Juan Capistrano, CA) 7264-2K accelerometers were mounted mutually orthogonal at the center of gravity of the headform. Three 12000 deg/sec angular rate sensors (DTS, Inc., Seal Beach, CA) were also mounted orthogonally at the center of gravity of the headform. A six-axis upper neck load cell (Denton ATD, Rochester Hills, MI) was

utilized to measured neck forces and moments. The data acquisition system for the HIII dummy was TDAS PRO (DTS, Inc., Seal Beach, CA). Data were collected at 10 KHz and processed according to SAE J211-1 (SAE 1995).
A Redlake HG 100K camera (Integrated Design Tools, Inc., Tallahassee, FL) was placed 1.5 m from center of the surrogate head at a lateral plane view. The camera was kept stationary throughout the testing. The surrogate was moved such that the dominant hand of the boxer was always closer to the camera. The video was collected at 500 frames per second.
The FAB system (Biosyn Systems, Surry, British Colombia, Canada) was used to analyze the motion of each boxer's punch. Anthropometric measurements of the participant were entered into FAB for accurate kinematic data. The FAB system consists of 13 wireless sensors, which contain accelerometers, gyroscopes, and magnetometers. The gyroscopes and accelerometers collect data at 100 Hz and the magnetometers collect data at 25 Hz. The sensors use internal processing and filtering to provide data output at 25 Hz. FAB is rated to collect acceleration up to 5 g and angular velocity up to 1200 degrees per second. The FAB is capable of collecting: foot sole pressure and weight; angle and position of each body segment (excluding the phalanges); and velocity and acceleration of body segments. It utilizes this data to calculate force, power, and torque of the cervical, trunk, shoulders, elbows, hips, and knees. For purposes of the current research, the force production of the lower body was examined.
Boxers were instrumented with FAB. Hand wraps and certified gloves were provided for all participants. The height of the HIII dummy was adjusted to the height of the boxer. Boxers were instructed to punch the dummy a total of four times—two times with a hook and two times with a straight—in their normal fashion. In this test, the hook punch was delivered with the dominant hand to the side of the dummy's head. A straight punch, also known as a cross, was delivered with the dominant-hand starting near the jaw of the boxer and travelling straight to the jaw of the recipient. The order of the punches was randomized between boxers. Boxers were instructed to use maximum effort for all four punches. Boxers had a total of 20 seconds to throw all punches. This allowed the boxers adequate time to recover after each punch and prepare for subsequent punches.
The data from the surrogate were processed using Diadem 11.0 (National Instruments, Austin, TX). Sensor offsets were removed from all channels and the data was filtered according to SAE J211-1. The angular rate data were converted to angular acceleration by calculating the derivative. Linear and angular resultant accelerations of the headform were calculated using the resultant of the three axes. The linear acceleration in each axis was multiplied by the mass of the headform (4.54 kg) to calculate force, and added to the upper neck force in the same axis. The resultant force (Punch Force) was calculated by the vector sum of the forces in the three axes.
TrackEye Motion Analysis (Photo-Sonics, Inc., Burbank, CA) 2.6 was used to measure pre-impact hand velocity. Marker tape was placed on the dorsal side of the distal forearm. This tape was used for 2-D video tracking. Each video was calibrated using marker tape of a known width (5.08 cm).
Data from the FAB system were used to calculate a Sum of Lower Body Forces (SLBF) for the dominant side of the body. The force output data from the FAB at the hips, knees, and feet at each data point were added to create the SLBF. The individual force values were added to form the SLBF because the maximum of these forces occurred at the same time value. Statistical analysis was conducted using PASW Statistics 18 software (SPSS, Chicago, IL). The data was divided into two groups by the type of punch (hook and straight). For both groups, correlations were conducted to compare punch force with SLBF and hand velocity. The data was analyzed using a p value of 0.05. If correlations were found to be significant, R^2 values were used to compare the correlations.

RESULTS: For each boxer, two punches were analyzed: the hook and straight with the greater punch force. This method ensured that the analyzed data most accurately represented the maximum effort of each boxer.

Correlations were conducted comparing SLBF and hand velocity to punch force for both punch types. All of the correlations were found to be statistically significant ($p < 0.05$). Figures 1 and 2 show the correlation between punch force and hand velocity for hook and straight punches respectively. R^2 = 0.380 and 0.391, respectively. Figures 3 and 4 show the correlation between punch force and SLBF for hook and straight punches, respectively. R^2 = 0.103 and 0.099, respectively. A summary of the R^2 and p values are contained in Table 1.

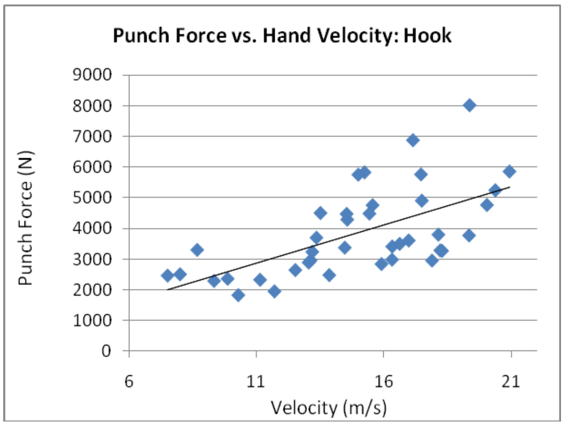

Figure 1. Correlation between punch force and hand velocity for hook punch.

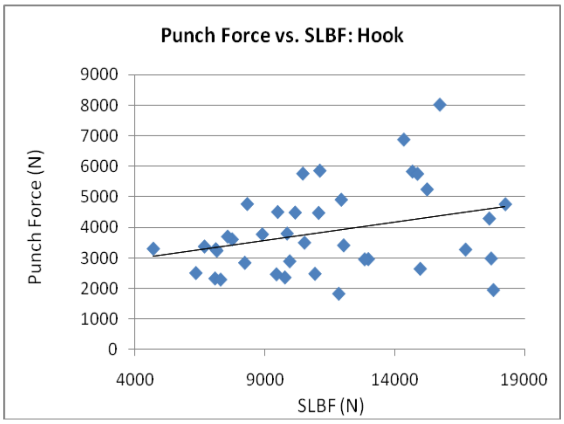

Figure 3. Correlation between punch force and SLBF for hook punch.

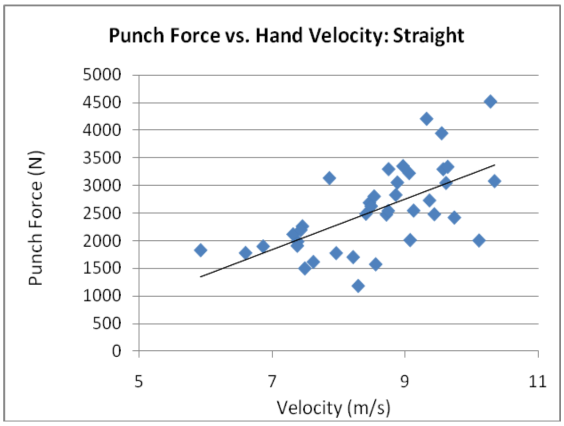

Figure 2. Correlation between punch force and hand velocity for straight punch.

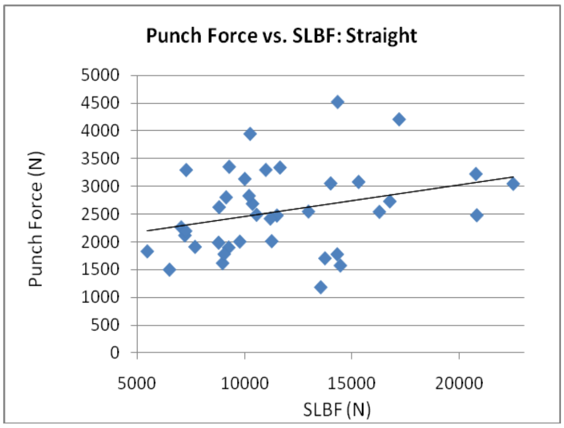

Figure 4. Correlation between punch force and SLBF for straight punch.

Table 1. Correlation Results

Variable	Hook		Straight	
	R^2	P	R^2	P
Hand Velocity	0.380	< 0.001	0.391	< 0.001
SLBF	0.103	0.043	0.099	0.048

DISCUSSION: This study evaluated the correlation between boxers' biomechanics and their punch force for two different punches. It was determined that punch force correlated more closely to hand velocity than to SLBF.
The FAB forces for the upper body were not used in this analysis because the maximum of these forces occurred at or near the time of the punch impact. Data within 80 ms of the event could not be analyzed due to a sensor-skin motion artefact apparent in visual observations and data evaluation. If the accelerometer range and the system sampling rate were increased the system could be greatly improved for impact conditions. Modifications to the

sensor attachment systems could limit sensor-skin motion artefact and allow upper body forces to be measured with the FAB.

The SLBF values generated by FAB did not correlate well to punch force. It is unknown if another data collection technique (force plates and multi-camera motion capture system, for example) could provide an improved correlation with punch force. It is possible that there is a limited correlation between lower body forces and punch force regardless of how the measurement is performed. Lower body forces may not be uniformly transferred to upper body forces due to differences in technique.

Although the FAB did not provide the best correlations, its uniqueness was intriguing to prospective test subjects. It proved invaluable for recruiting participants to the study. Many participants joined the study because of their interest in the FAB system.

CONCLUSION: These data indicate that the pre-impact velocity of a boxer's punch provides a more reliable indication of punch force than the pre-impact forces generated by a boxer's lower body as measured by FAB. Punch force correlates to hand velocity with an R^2 value of 0.380 and 0.391 for hook and straight punches, respectively. However, punch force correlates to SLBF with an R^2 value of 0.103 and 0.099, respectively. Boxers and boxing trainers may benefit from an understanding of these correlations to improve performance.

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Oral Session 20

Kicking

KICK IMPACT CHARACTERISTICS FOR DIFFERENT RUGBY LEAGUE KICKS

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Kicking is becoming increasingly important in rugby league and one of the most important aspects of kicking is the nature of impact. This study examined impact characteristics for five rugby league kicks – the goalshot, the punt kick, the dropkick, the grubber and the 'bomb'. Seven elite players performed these kicks while being videoed (6000Hz). Digitised data of nine points (five on the kick leg, four on the ball) were used to quantify parameters near and during impact. Rugby league kicks produced a smaller time in contact and a larger amount of work compared to kicks in other sports. Differences also existed between the five types of rugby league kick tested in terms of ball to foot ratio and foot and ball positioning at impact. These differences between kicks within the same sport highlight the need to evaluate different kick types separately within a sport.

KEYWORDS: Ball to foot**,** Drop kick, Goalshot,

INTRODUCTION: Kicking has become increasingly important in rugby league. Where once the kick was used to get out of defense and attacking kicks were predominantly 'bombs' (a high trajectory into the attacking scoring zone that gave attacking players time to get under it) recent years has seen the emergence of more precise kicks to wide receivers across the ground. The kicking game of National Rugby League team the Melbourne Storm is considered to be a key reason for their success in gaining four grand finals in a row and in 2007, more than a third of their tries were scored from kicks (Ball, 2007). The advent of the golden point rule, where the first team to score in extra time (played when scores are tied at the end of regular time) has seen field goal kicking once again becoming important.

An important part of any kick is impact between the foot and the ball. This is the only point at which the performer can propel the ball during the kick and so factors at impact must be influential to ball trajectory. Ball to foot factors have been found to be important in soccer kicking (e.g. Nunome et al., 2006) and in Australian football for kick distance (Ball, 2008a) and between the preferred and non-preferred foot (Smith et al., 2009). Further, in Australian Football (AF) where kicking is the main skill in the game, impact has been deemed the most important aspect of kicking in coaching resources (AFL kicking committee, 2009).

To date, no study has examined impact characteristics for rugby league kicks. Given the different balls used for rugby league compared to soccer and AF, these characteristics might be expected to differ. There are obvious differences in shape between the soccer ball (spherical) and the league ball (ovoid). The league and AF balls are more similar in shape, but the league ball is synthetic rather than leather, is slightly lighter (approximately 400g compared to the 450g of an AF ball) and has a slightly rounder overall shape. As such, ball to foot characteristics might be different and need to be evaluated specific to rugby league.

A limitation of previous impact studies is the small number of kick types that have been examined (AF: drop punt only; soccer: instep and toe kicks). In AF for example, kicks include spirals (where the ball rotates about its long axis in flight) and snaps (where the ball spins about its short axis) in addition to the drop punt examined in impact studies. Similarly, many variations of the instep or toe kick are evident in soccer. Useful information on the nature of foot to ball impact can be gained from examining these different kicks within each game.

The aim of this study was to examine impact characteristics of rugby league kicks and to determine if these characteristics differ for different kicks.

METHOD: Seven elite rugby league players contracted to an Australian National Rugby League performed kicks using a Steeden rugby league football (used in NRL competition). All players performed a drop punt over 45 m, a 15 m grubber kick (kick that rolls along the ground), a 'bomb' (an attacking kick that requires maximum or near maximum height), a drop

kick at goal from 40 m from the goal and goalshots from a placekick. These were chosen as they are found to be the most frequently used kicks in the National Rugby League competition (Ball, 2007). Players only performed kicks in testing if they were currently performing them in games (hence N varied for different kicks, see table 1). This was also the reason for not including the kick off, line dropout and kick for touch from a penalty which are all prevalent in the game but only one player had performed them in games during the current season (note: this is typical of most clubs). With the exception of the goalshot for which a minimum of four kicks was performed, players were allowed to kick up to five of each kick until they had achieved what they, and their kicking coach who attended testing, felt was a good kick. A Photron Fastcam APX-RS high speed camera (Photron Ltd, San Diego) operating at 6000Hz was placed perpendicular to the line of kick. The field of view was maximised for digitising accuracy so was zoomed in to capture the ball, foot and kick leg knee for a minimum of ten frames before until ten frames after ball contact.
Eleven body and boot landmarks were digitised using Silicon Coach Analysis tools (Silicon Coach Ltd, NZ) for the ten frames before and after ball contact. Data were then transferred to Microsoft Excel to calculate foot speed (ten frames prior to impact, average of the ankle and three foot markers), ball speed (ten frames after ball and foot parted, average of four points on the ball, the top point, bottom point, two mid points), ball:foot speed ratio, time in contact (from initial contact to the point of separation of the ball and foot), ball displacement (displacement from initial contact to separation point) and work (W) = mad [where m (of ball) = 0.401 kg, a (of ball) = change in ball velocity from before contact to after separation divided by contact time between foot and ball, d = ball displacement while ball in contact with the foot, Ball 2008b]. The change in shank angle (difference between the angle of the shank, defined by the head of the fibula and lateral malleolus, from impact to separation) was also quantified but due to the lateral body lean with which the goalshot and drop kick are performed, this parameter was not quantified for these kicks.

RESULTS: Table 1 reports mean values for the five kicks assessed.

Table 1. Mean values for five rugby league kicks

		45m (N=7)		Bomb (N=4)		Field Goal (N=5)		Grubber (N=6)		Goalshot (N=4)	
		m	sd	m	sd	m	sd	m	sd	m	sd
Foot Speed (m/s)	Before BC	20.0	2.5	21.0	1.5	21.8	1.6	11.0	2.3	21.2	1.7
Ball Speed (m/s)	After BC	25.8	1.8	26.9	4.2	26.5	2.4	13.7	2.4	25.2	4.0
Ball:Foot Ratio		1.30	0.13	1.28	0.12	1.22	0.11	1.27	0.25	1.20	0.20
Time in Contact (ms)	During BC	7.2	0.6	6.8	0.2	7.1	0.5	8.8	0.7	7.4	0.3
Ball Displacement (m)	During BC	0.20	0.02	0.20	0.01	0.23	0.04	0.12	0.03	0.22	0.02
Work (J)	During BC	290	49	342	50	316	48	68	17	306	71
Ball Angle (°)	Before BC	66	10	75	5	70	4	12	9	36	4
	After BC	70	8	77	9	65	17	20	25	53	8
	Change	4	8	2	10	5	17	-8	19	17	4
Foot Trajectory (°)	Before BC	7	5	24	13	8	7	0	8	2	5
Ball Trajectory (°)	After BC	32	11	53	12	39	9	1	2	36	3
Support Foot to	Horizontal	0.39	0.15	0.58	0.07	0.23	0.15	0.60	0.18	0.15	0.14
Ball distance(m)	Vertical	0.40	0.07	0.63	0.05	0.30	0.10	0.23	0.04	0.28	0.01

DISCUSSION: The aims of this study were to provide data for impact characteristic data for rugby league kicks and to compare different kicks. There is no comparison data in the scientific literature for rugby league kicking. Ball speeds and angle of trajectories were similar to values reported by Holmes et al (2006) for rugby union kickers (goal kick = 26.44 m/s, 30.22°; drop kick = 25.60 m/s, 35.76°). Ball displacements during foot contact of 0.20-0.23m for the four long kicks in this study lay between values reported for AF (0.19m and 0.24 m for 30m and 50m kicks, Ball, 2008b; 0.19 m and 0.23 m for preferred and non-preferred kicks, Smith et al., 2009) and soccer (0.15 m, Asai et al., 2002; 0.26 m Tsaousidis and Zatsiorsky, 1996). Ball speed to foot speed ratios also lay between values reported for soccer (1.1, Asai et al., 2002; 1.37 Nunome et al., 2002) although the 50 m kick, bomb and grubber were all slightly higher than the 1.23 reported for preferred foot kicks for AF (Smith et al., 2009).

Differences in impact characteristics existed between rugby league kicks and those of other sports. Time in contact values for all kicks (6.8 ms for the bomb up to 8.8 ms for the grubber) were lower than any other values reported (AF: 9.8-10 ms, Ball, 2008b; 11.5ms, Smith et al., 2009; soccer: 9ms, Asai et al., 2002; 16ms, Tsaousidis and Zatsiorsky 1996; 11.1 ms, Nunome et al., 2006). The smaller mass of the rugby league ball is likely to have contributed to this difference (rugby league ball = 0.400kg, soccer ball = 0.435 kg, AF ball = 0.450 kg). For the same force, the rugby league ball will experience greater acceleration compared with the soccer and AF balls. Certainly, the difference in contact time cannot be explained by kick intensity. While for most studies, kicks were maximal (compared to 45-50m kicks in this study, or 80-95% maximum), Ball (2008b) performed kicks over a shorter distance and contact time remained larger than for rugby league kicks (30 m kick, 9.8 ms contact time). Further, the shortest kick for this testing (grubber) also exhibited the largest contact time.

Work on the ball for all kicks excluding the grubber was larger than values reported for AF kicks (225-271 J, Ball, 2008, Smith et al., 2009). It was also larger for the 45m kick and bomb compared to soccer kicks (290-310 J, Tsauosidis and Zatsiorski, 1996) and values were similar for the 45 m kick (290J) and goalkick (306J). Of note here is that the kicks in this study were sub-maximal while the soccer kicks were maximal. Given Ball (2008) reported work increased with increased kick distance, it is possible that maximal goalkicks and drop kicks might exceed soccer kicks for work done on the ball. Coefficient of restitution (COR) is a likely contributor to this finding although no data exists for the rugby league ball for comparison. Where COR has been evaluated under kicking conditions, rugby balls (0.77-0.81, Gallagher and Cooke, 1998) had much higher values than soccer balls (0.45-0.65, Dorge et al., 2002). Given the similarities between league and union balls, a similar COR might be expected. The differences in mass might also contribute. Once again, the sub-maximal kick performed in this study cannot explain this finding. Ball (2008) reported larger work values for longer (or more maximal) kicks so the submaximal kicks in this study would be expected to produce smaller and not larger work values. Further, had the field goal and goalshot been maximal, it is likely they would have also produced more work than soccer kicks.

Differences existed between the rugby league kicks examined in this study. Not surprisingly, the grubber was significantly different to other rugby league kicks for most parameters due to its trajectory being along the ground and its aim to stop within 15 m. Further, it was noted from video footage than a large range of techniques and ball angles were used for this kick. While foot speeds and trajectories (foot and ball) were similar, players angled the ball differently to obtain different spin and run along the ground. This variance was evident in large standard deviation values for ball angles before and after contact in table 1.

Differences also existed between the longer rugby league kicks examined in this study. Ball to foot speed ratios ranged from 1.2 to 1.3 (8% difference) and work ranged from 290 J to 342 J (18% difference). Other differences related to ball and foot position/trajectory also existed with ball angle, ball and foot trajectory and support foot position differing between a number of the kicks. These ‘position’ factors were related to the kick task. For example, during the bomb, for which maximum height is desirable, the ball was higher from the ground and foot trajectory was more vertical at ball contact compared to the other kicks for which horizontal distance was the more dominant aspect of the task.

With the small N, statistical analysis was not appropriate so future testing with larger numbers is needed to determine if the differences between parameters is significant. However given the large relative differences that existed between kicks in this study it is clear they need to be evaluated separately for impact characteristics. These differences probably also exist between kicks in other sports. Currently soccer kick impact analysis has largely

been limited to instep and toe kicks and only one kick (the drop punt) has been examined in AF. As such there would seem to be a need to evaluate more kick types in these sports.
Ball orientation and change in ball angle during contact produced interesting results. For the goalshot the ball moved through 17 degrees from the instant of foot contact until the instant of foot-ball separation. This was less evident in the other kicks with the field goal, bomb and 50m kick only moving through 5 degrees. The point of contact on the ball and the orientation of the ball at contact would have influenced this. However it was also noted that among the goalkickers that the most junior goalkicker was far more variable (range of values 8 – 19 degrees) compared to the more senior kickers (range of values less than 2 degrees for all three kickers). The consistency of ball strike and how it relates to performance is an important future direction for this work, as is the evaluation of how this large change in angle influenced performance compared to field kicks.

CONCLUSION: This study has provided impact data for a range of different rugby league kicks. In comparison to AF and soccer kicks, rugby league kicks exhibited smaller time in contact. A larger amount of work done on the ball compared to AF kicks while showing similarities for foot and ball speeds, ball to foot ratio and displacement of the ball while in contact with the foot. Further, differences existed between the different types of rugby league kicks tested, indicating these kicks need to be evaluated separately. This is also likely to generalise to different kicks within other sports such as AF and soccer so evaluation of different types of kick is necessary in these sports.

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THE SUCCESS OF A SOCCER KICK DEPENDS ON RUN UP DECELERATION

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The purpose of the study was to relate the motion of the centre of mass (CoM) during run ups in soccer full insteps kicks with the obtained ball speeds. Nineteen experienced players performed kicks onto the goal and their full body kinematics as well as ball motion were analysed in three dimensions using two high speed video cameras. Higher decelerations of the CoM with the last step are associated with higher ball velocities and higher thigh angular impulses. Those data suggest, that an intensive breaking of the CoM velocity provides a prerequisite to transfer a portion of the CoM impulse into angular impulse of the thigh. High angular impulse of the thigh however can be beneficial for fast instep kicks.

INTRODUCTION: It is widely accepted and quite easy to understand, that the ball speed of a soccer kick depends on the action of the kicking leg (Tsaousidis & Zatsiorsky, 1996; Lees & Nolan, 1998; Sterzing & Hennig, 2008). Both the swing phase prior to ball impact and the collision phase determine the quality of a soccer kick (Tsaousidis & Zatsiorsky, 1996). A high knee muscle moment and a high knee interaction moment at the kicking leg appear to be beneficial to achieve high ball speeds (Nunome et al., 2006a; 2006b). Little attention however has been drawn to the influence of the supporting leg on ball speed. It is reported, that higher ball speeds are associated with higher peak ground reaction force (GRF) values (Barfield, 1998; Orloff et al., 2008) but mechanisms how reaction forces acting on the player's centre of mass (CoM) lead to higher ball speeds are not understood. Findings from throwing disciplines suggest, that an intensive deceleration of the CoM combined with a transfer of momentum and energy from lower body parts to the arm is a prerequisite for successful throws (Morriss et al., 2001; Morriss & Bartlett, 1996). It has been suggested, that the interaction of the supporting leg with the surface has an influence on kicking performance (Potthast & Brüggemann, 2010; Sterzing & Hennig, 2008). Therefore the purpose of this study was to investigate if the deceleration of the player's CoM during the last step is related to ball speed in full instep kicking.

METHODS: 19 experienced male soccer players (highest German amateur level) performed 5 shots from a central position at the end of the penalty box onto the goal. The upper half of the goal was subdivided into three areas. Left footed players had to aim for the upper left corner, right footed players for the upper right corner. The subjects were instructed to perform full instep kicks after self selected run up. The run up and kicking kinematics of the players were recorded by two synchronized digital high-speed cameras (Basler, 100 Hz). The cameras recorded from a posterior and from a lateral view (figure 1). Video footages were analysed (Vicon Motus 9.2) by identifying anatomical landmarks of the player (head, C7, left and right shoulder, elbow, hand, hip, knee, ankle, heel, tiptoes) as well as the centre of the ball. Next to the segmental kinematics the movement of the CoM was calculated. Segment kinetics for the swinging leg were calculated using segmental inertial properties and kinematics. Particularly the angular impulse was calculated by multiplying the moment of inertia and the angular velocity component-by-component. To assess kicking performance the ball speed was used by analysing the ball movement ten frames after foot ball contact. For each player the fastest kick hitting the target area was considered for further analysis. The deceleration impulse during the last step with the supporting leg was calculated by multiplying the players' mass with the change of run up velocity during stance. All kicks were performed on one artificial turf system (pure rubber infill). This pitch did not change the kicking motion of the players in comparison to a well maintained natural grass pitch (Potthast & Brueggemann, 2010).

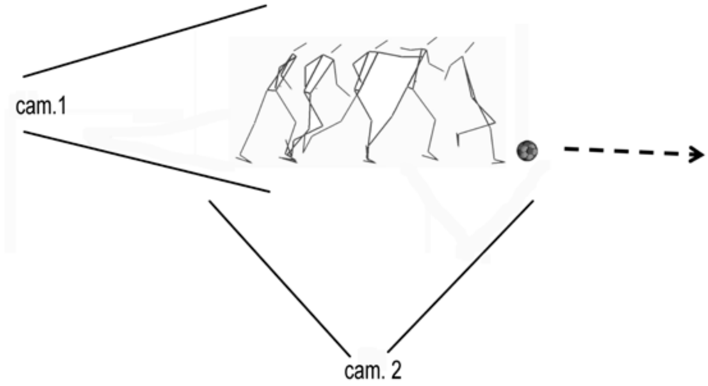

Figure 1. Schematic drawing of the camera positions (cam. 1 and cam 2) in respect to the players' run up and kicking action. The dashed arrow indicates the kicking direction.

RESULTS: The average kicking velocity was 100.1 km/h ± 7.3, the average deceleration impulse was 144.5 kg*m/s ± 32.5, the average increase of thigh angular impulse was 3.2 kg*m^2/s ± 1.1. Figure 2 shows the relationship between kicking velocity and deceleration impulse. The correlation coefficient was r=0.6 (p=0.006, n=19), indicating that 36% of the variance of the kicking velocity is explained by the deceleration impulse of the CoM. More intensive reduction of the velocity of the CoM is correlated with higher ball speeds.

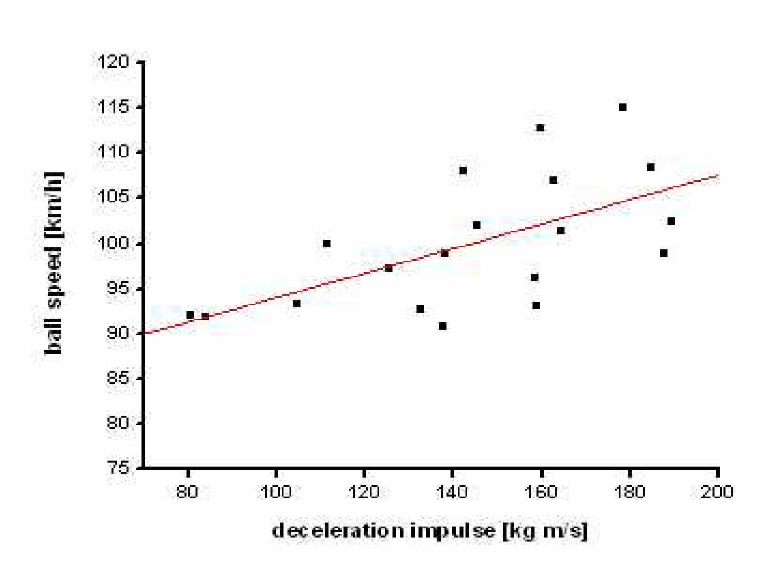

Figure 2.Relationship between CoM deceleration impulse with the last step of the supporting leg and ball speed (r=0.6; p=0.006; n=19). A more intensive breaking is associated with higher kicking velocities.

Figure 3 indicates, that a bigger portion of the CoM deceleration impulse is transferred into angular impulse of the thigh in kicks with high ball speed than in kicks with low ball speed.

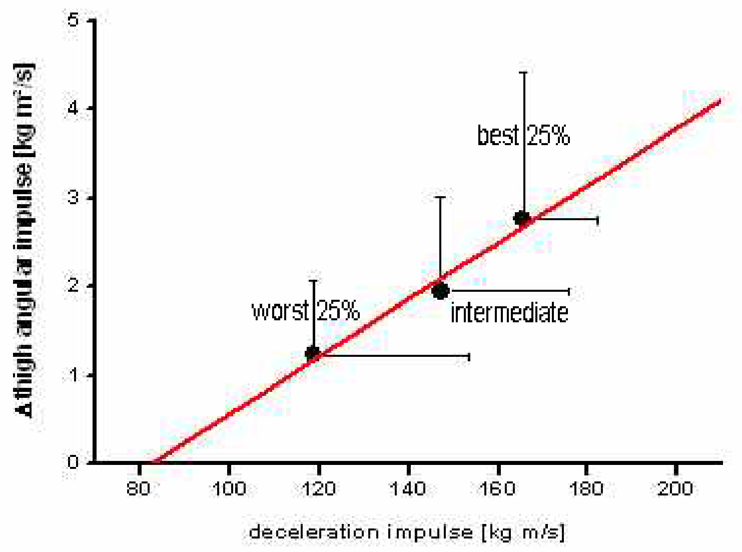

Figure 3. Mean values and standard deviations of the angular impulse (vertical axis) and decelereation impulse of the CoM (horizontal axis) for kicks with high ball speeds (best 25%), low ball speeds (worst 25%) and intermediate ball speeds (n=19).

DISCUSSION: The purpose of the study was generally to investigate if a momentum transfer to leg segments due to the deceleration of the run up velocity of the CoM could influence ball speed. Using video footages three-dimensional run up and kicking kinematics were analysed as well as ball speed. The ball speed parameters were similar to those reported in other studies. Slight differences (Sterzing & Hennig, 2008) can be explained by different measuring techniques and skill level of subjects. The results show, that intensive deceleration of the run up within the last step of the stance leg is correlated (r=0.6) to higher ball speeds. In addition kicks with the highest ball speeds coincide with high CoM decelerations and high increases of angular impulses of the thigh. On the other side, the slowest kicks coincide with smaller CoM decelerations and smaller changes of thigh angular impulses. Those facts indicate, that an intensive breaking of the CoM velocity provides a prerequisite to transfer a portion of the CoM impulse into angular impulse of the thigh. High angular impulse of the thigh can be beneficial for fast instep kicks (Nunome et al., 2006a, 2006b). This information should be valuable in different fields of application. The results indicate that the deceleration of the run up and the behaviour of the stance leg in general have to be considered when teaching in-step kicks. In the field of sports technology an un-disturbed deceleration has to be ensured e.g. by surfaces or shoes. It should be sated that this study needs confirmation by future research.

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COORDINATION PATTERNS OF PREFERRED AND NON-PREFERRED KICKING OF THE DROP PUNT KICK: A KINEMATIC ANALYSIS OF THE PELVIS, HIP AND KNEE

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This study expanded on previous work investigating preferred (P) and non-preferred (NP) leg kicks in Australian football (AF), however this work included the kinematics of the pelvis and hip as well as the knee. Eight elite AF players performed drop punt kicks with their P and NP legs. Three dimensional kinematic data (Optotrak Certus, 250Hz) from kick foot toe off to ball contact was recorded for each kick. Significantly larger foot speed and knee angular velocities were produced by the P leg. Differences in coordination were found to be largely ROM and velocity based. Of note was the P leg produced greater ROM at the knee and pelvis but the NP leg produced greater ROM at the hip suggesting a different strategy might exist for the different legs. More work exploring the different involvement of the hip, knee and pelvis is warranted.

KEYWORDS: Coordination, kicking, preferred and non-preferred leg.

INTRODUCTION: Australian Football (AF) is a popular Australian team sport. The aim of the game is to progress the ball down a field and score as many goals and points as possible by kicking the ball between posts at the attacking end of the ground while at the same time, preventing your opponents from scoring. Ball progression can be achieved by kicking or handballing the ball to a teammate or to space and kicking is the only method of scoring goals. Due to the high pressured nature of the game, AF requires its players to kick with their preferred leg (P) and non-preferred leg (NP). Ball (2003) reported that 20% of kicks in AF are performed with the NP and this can be as high as 45% for some players. Players who can kick equally well on either leg are seen to have a tactical advantage as they can produce more rapid and successful ball disposals.

Only one study has examined the kinematic differences between P and NP kicking in AF although low N limited statistical analyses. Hancock and Ball (2008) found non-significant large effects for the difference in knee ROM between the two kicking legs at (P = 0.12, d = 0.9). The study also examined phase plane diagrams of knee angle-knee angular velocity and reported similar overall shapes of the curves but the non-preferred foot curve was smaller than that of the preferred foot curve. The authors concluded that differences between the P and NP were range of motion and angular velocity based and not coordination or timing for elite performers.

A limitation of the Hancock and Ball (2008) study was that it was limited to the knee. To more fully understand differences between the P and NP kicking, more joints and segments need to be examined. In particular, the hip joint and pelvis segment have been shown to be important in kicking and need to be included in this analysis. Dichiera et al. (2006) reported differences in pelvic motion and positions between accurate and inaccurate AF kickers and substantially larger joint moments have been reported at the hip (220Nm) compared to the knee (90 Nm) during the kick suggesting the hip is a principal area for force generation (Robertson and Mosher, 1985). As such, it is expedient to expand the exploration of coordination differences in the AF to include the hip and pelvis, a point that Hancock and Ball (2008) make.

The aim of this study was to determine if differences exist between preferred and non-preferred leg kicks for kinematic factors and coordination, and including evaluation of the hip and pelvis.

METHOD: Eight elite Australian Rules football players, currently playing in the Australian Football League (AFL) were recruited for this study. All players reported being right foot dominant and free of any lower limb injury. Details of participants are shown in Table 1.

Table 1. Participant Details

	Age (years)	Height (cm)	Mass (kg)	Years on AFL List
Participants	21.3 ± 0.9	1.87 ± 0.15	85 ± 5	3.6 ± 2.1

Participants performed 10 test kicks using new Sherrin AF footballs inflated within the specified pressure range of 62-76kPa (Australian Football League, 2003). Five kicks were performed using their P and five kicks were performed using their NP. Participants used their preferred run up and kicked into a net 5 m away (directly in front). Participants wore standard training apparel (football jumper, shorts) as well as their regular training shoes ('turf' training shoes).
During each kick, 250 Hz three dimensional data of the shank, thigh, knee, hip and pelvis were collected using Optotrak Certus System (Northern Digital Inc., Waterloo, Canada, root mean square error of 0.1mm in 2D and 0.15mm 3D). Three Optotrak cameras were placed approximately 5 m from the kick area in three different positions (behind the kick, to the left of the kick, to the right of the kick) directly facing the designated kicking zone. Clusters of infra-red light emitting diodes (LED) were placed on the pelvis, shank, thigh and foot of both legs of each participant. The thigh clusters were placed on the lateral side of the thigh midway between the greater trochanter and the lateral epicondyles of the femur. The shank markers were placed on the lateral side of the shank, midway between the lateral malleolous and the distal end of the calf muscle. The pelvis LEDs were positioned relative to the anatomical landmarks (AL's) of the posterior superior iliac spine and the sacrum. An additional single marker was placed on the shoe over the head of the fifth metatarsal. AL's were located and established at the pelvis (left/right iliac crests and posterior/ superior iliac spines), hip (left/right greater trochanter), knee (medial/lateral epicondyl) and ankle (medial/lateral malleolus) using a digitising probe (Northern Digital Inc., Canada).
The P and NP kicks were compared for coordination profiles and kinematic data identified as important in previous research. Coordination profiles were evaluated using angle-angle diagrams (knee-hip angle, hip-pelvis angle) and phase plane diagrams (knee angle-knee angular velocity). Kinematic data included foot speed, pelvis, hip, thigh, knee and shank angles, angular velocities and ranges of motion (all identified as important in AF kicking, Ball, 2008). T-tests were conducted for each mean value to determine the statistical significance of each value. Angles were 2D in the saggital plane as the AF kicking movement is planar. The alpha level was set at $p < 0.05$ for significance. Effect sizes were additionally calculated (Cohen's *d;* small *d* = 0.2, medium *d* = 0.5, large *d* = 0.8, Cohen, 1988).

RESULTS: Table 2 reports significant differences between preferred and non-preferred legs.

Table 2. Significant differences between the preferred and non-preferred leg kicks.

	Foot Speed (m/s)		Knee Angular Velocity at Ball contact (°/s)	
	P	**NP**	**P**	**NP**
Mean	19.1	17.2	1302	1043
SD	1.5	1.5	216	176
P-Value	0.05		0.04	
Effect Size	1.3		1.3	
Effect Scale	Large		Large	

Non-significant large effect sizes also existed for knee range of motion (*d* = 1.24), hip angular velocity at ball contact (*d* = 0.71) and hip range of motion (*d* = 1.20). No significant differences or medium or large effects were found for timing parameters.

Figure 1 shows the mean hip-pelvis angle-angle cyclogram with a difference near BC noted (shaded area).

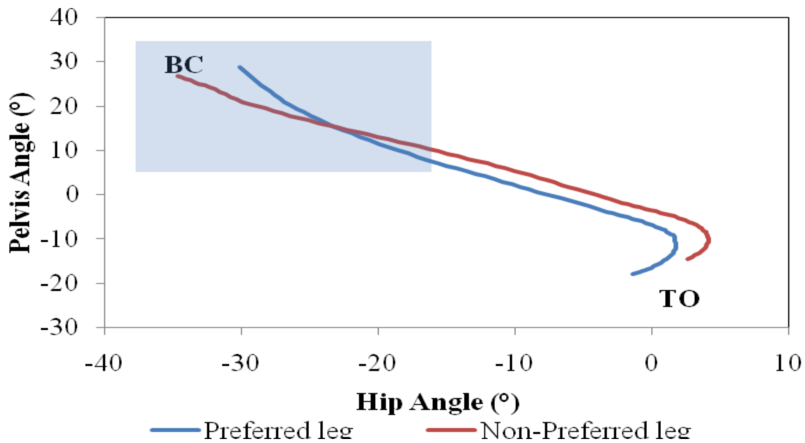

Figure 1. Mean Hip-Pelvis Angle-Angle Cyclogram for the Preferred and Non-Preferred Kicking Leg (TO= Toe Off, BC = Ball Contact)

DISCUSSION: The significant difference in foot speed between P and NP in the current study supported previous research. Hancock and Ball (2008) study found a large non-significant effect existed between the two kicking legs (*P* = 0.09, *d* = 1.5) and with the same N as this study, this would have been a significant result (note the larger effect size). Similar findings have been reported for soccer kicking with both Dorge *et al*., (2002) and Nunome *et al*., (2006) reporting significant differences in foot speed between the P and NP kicks. A large significant effect was found to exist for knee angular velocity at ball contact for the preferred kicking leg (P = 1302 °/s, NP = 1043°/s). The difference found in this study indicated that elite players can generate more foot speed and knee angular velocity with their preferred leg.

Angle-angle diagrams and non-significant findings for timing parameters indicated that differences between the P and NP were range of motion and velocity rather than coordination based. This supported the findings of Hancock and Ball (2008) but with the inclusion of more joints in this study. Based on these findings, coaching should focus on increasing range of motion and foot speed of the non-preferred leg to improve performance.

Although ROM values were not significant, an interesting pattern emerged that is worth noting for future studies. ROM was larger for the P at the knee and pelvis but was smaller at the hip (medium or large effects existed for each comparison). This might suggest a different strategy or difference in control of movement between the preferred and non-preferred legs might exist. Future work exploring this mechanism is warranted.

The pelvis-hip angle-angle diagram in figure 1 showed interesting results. The shaded area illustrates a subtle difference in the movement pattern just prior to ball contact. There are two possible explanations for this difference. The hip angular velocity of the P could have decreased more than the NP in preparation of ball contact. However, as the pelvis angle for both legs

reached similar values at ball contact it is more likely that the pelvis angular velocity of the P increased in order to generate more power for ball contact.
During performance of the NP, participants could have also placed constraints on the pelvis and knee ROM in order to simplify the control of the movement by limiting it to the hip joint. This explanation can be linked to Bernstein's (1967) theory of locking degrees of freedom. The hip for the non-preferred leg might be easier to control and therefore, the player locks down (or restricts at least) the surrounding joints to allow for more movement to be conducted at the selected joint or segment. Conversely, hip angular velocity of the preferred leg decreased (48°/s) toward the point of ball contact while the pelvis angular velocity continued to increase (124°/s). The pelvis could have been used to generate more of the power at ball contact. As such, rather than a larger hip extension, the larger last stride might be related to a greater pelvis ROM which in turn leads to a larger pelvis angular velocity at ball contact.

CONCLUSION: Differences existed between the preferred and non-preferred foot in AF kicking. Significantly larger foot speeds and knee angular velocities were produced by the preferred foot kicks. Observation of angle-angle diagrams indicated that differences were ROM rather than timing and coordination based. However some interesting differences existed with ROM being larger at the knee and pelvis but smaller at the hip for the preferred foot. Future work with large N is warranted to further explore this aspect of AF kicking.

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KICK IMPACT CHARACTERISTICS OF JUNIOR KICKERS

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Impact is important to kicking performance and while differences have been found between kick distances and between preferred and non-preferred leg kicks, no work has examined junior kickers. This study examined impact characteristics of the Australian football (AF) drop punt kick for juniors and compared these data with seniors from Smith et al. (2009). Twenty one junior AF players performed a maximum distance kick. The foot, ball and shank were digitised from 6000 Hz video to calculate seven foot/ball parameters. Junior players produced significantly smaller foot and ball speeds but not foot to ball speed ratios compared to senior players. Work was also significantly different due to lesser force being applied to the ball. Junior players should focus on increasing foot speed and force on the ball to increase kick distance.

KEYWORDS: Australian football, junior skill development, punt kick

INTRODUCTION: The nature of impact between the foot and the ball is an important technical factor in the kicking skill. Ball velocity in soccer kicking has been proposed to be developed by a combination of foot speed and the nature of impact between foot and ball. Similarly, in Australian Football (AF), it has been suggested that kick distance is influenced by the nature of impact between the foot and ball (Baker and Ball, 1996). Further, the importance of impact has been highlighted by recent recommendations made by the Australian Football League (AFL) kicking committee that coaches should prioritise evaluating impact before other technical aspects.

An important influence on the characteristics of impact is footedness. Impact factors have been shown to differ between preferred and non-preferred foot kicks in AF and soccer. Smith et al. (2009) compared maximal distance kicking for the preferred and non-preferred foot kicking for elite AFL players from 6000 Hz video. Smith et al. (2009) found that foot speed, ball speed, change in shank angle and work differed significantly between feet (see table 1 in results). However, no significant difference existed for time in contact with the ball or for the ratio between foot speed and ball speed. Similarly, foot and ball speed did but foot to ball speed ratio did not differ significantly by foot in soccer (Nunome et al., 2006a, preferred = 1.35, non-preferred = 1.32).

Impact factors have also been shown to differ for kicks of different distances. Ball (2008a) examined impact in 1000Hz video of 30 m and 50 m kicks, performed by eight elite Australian Football League (AFL) players. Significant differences existed between 30 m and 50 m kicks for change in ball velocity (50 m kick = 25.0 m/s, 30 m kick = 22.1m/s), change in shank angle during ball contact (50 m kick = 18 degrees, 30 m kick = 14 degrees) and work done on the ball (50 m kick = 271 J, 30 m kick = 198 J). The combination of significant change in ball velocity but not time in contact (50 m kick = 10 ms, 30 m kick = 9.8 ms) led Ball to conclude that the amount of force, rather than how long the force was applied to the ball was the key performance determinant for kick distance. This was supported by Smith et al. (2009) who found the preferred leg kick produced significantly larger change in velocity but with no difference in time in contact compared to the non-preferred foot (the preferred foot produced greater ball speed which is associated with kick distance, Ball 2008b).

An important extension of these previous examinations of the nature of impact is to determine what characteristics junior players exhibit and how these compare to senior performers. This information can potentially identify key parameters that differ between these age groups and more precisely guide programmes aimed at junior kicking development. As yet, few studies in any kicking sport have examined junior players and how they differ from senior players. The aim of this study was to measure impact characteristics for junior AF players and to compare them with senior players.

METHOD: Twenty-one junior AF (Age 16.9 ± 1.1 years; height = 1.78 ± 0.20 m; mass = 71.3 ± 8.1 kg) players participated in this study. To allow for direct comparison with senior data, the test protocols and analysis methods used by Smith et al. (2009) were used for this study. All players kicked a Sherrin Australian Rules football (used in AFL competition, pressure range of 67-75 kPa) for maximal distance with the preferred and non-preferred leg. Players were allowed to kick up to five kicks on each leg until they had achieved what they felt was a good kick. This kick was confirmed by their kicking coach who was in attendance. A Photron Fastcam APX-RS high speed camera (Photron Ltd, San Diego) operating at 6000Hz was placed perpendicular to the line of kick and was zoomed in to capture the ball, foot and kick leg knee for a minimum of ten frames before until ten frames after ball contact. Seven body/boot landmarks (head of fibula, lateral malleolus, heel of boot, head of the 5th metatarsal, toe of boot, top point of ball, bottom point of ball) were digitised for these ten frames before and after ball contact using Silicon Coach Analysis tools (Silicon Coach Ltd, NZ). Data was then transferred to Microsoft Excel to calculate foot speed (ten frames prior to impact, average of the ankle and three foot markers), ball speed (ten frames after ball and foot parted, average of the top and bottom of the ball), ball:foot speed ratio, time in contact (from initial contact to the point of separation of the ball and foot), ball displacement (displacement from initial contact to separation point), change in shank angle (difference between the angle of the shank, defined by the head of the fibula and lateral malleolus, from impact to separation) and work [calculated as m.a.d, where m (of ball) = 0.45 kg, a (of ball) = change in ball velocity from before contact to after separation divided by the time in contact between ball and foot, d = ball displacement while ball in contact with the foot, Ball 2008a). The force component of the work equation (m.a) was also used in evaluation of findings.
To compare with senior players, preferred leg kicking data from Smith et al. (2009) were used (N=18 senior AFL players, Age: 22.8 ± 4.2 years).
Paired t-tests were conducted to compare junior and senior players for each kicking parameter. Statistical significance was set at $p < 0.007$ (using a Bonferroni adjustment of $p < 0.05$ for seven parameters). Effect sizes (large: $d > 1.2$, medium: $d > 0.6$, small: $d > 0.2$) as defined by Cohen (1988) were also calculated for each comparison.

RESULTS: Table 1 reports mean values for the junior kickers along with comparison data from Smith et al. (2009). Statistical analyses are also included.

Table 1. Impact characteristics for preferred foot kicking for junior AF players. Comparison data from Smith et al. (2009, preferred leg kicking data only) and statistical analyses also included.

	Junior (N=21)	Senior (N=18) (Smith et al., 2009)	t-test (*p*-value)	Effect size (*d*)
Foot speed (m/s)	21.3 ± 1.3	26.5 ± 2.5	<0.001*	2.6
Ball speed (m/s)	24.7 ± 2.1	32.6 ± 4.4	<0.001*	2.3
Ball:foot speed ratio	1.16 ± 0.08	1.23 ± 0.11	0.02	0.7
Time in contact (ms)	11.06 ± 0.93	11.53 ± 1.25	0.19	0.4
Ball displacement (m)	0.20 ± 0.02	0.22 ± 0.02	<0.001*	1.0
Change in shank angle (°)	14 ± 2	13 ± 1	0.01	0.6
Work done on the ball (J)	135.7 ± 22.5	225.0 ± 45.0	<0.001*	2.5

* Significant difference (p<0.007) after Bonferroni adjustment.

Junior AF players produced significantly lower foot speed, ball speed, ball displacement and work values. These were large effects. Ball to foot speed ratio, change in shank angle and time in contact produced non-significant, medium effects.

DISCUSSION: The aim of this study was to measure impact characteristics in kicking for junior AF players and to compare with senior players. There is no comparison data for junior AF players and very limited data for comparison with junior kickers from other sports. Ball speeds for juniors in this study (16.9 years, 24.7 m/s) lay between previously reported values reported for maximal soccer kicking for juniors (10-17 years, 14.9-22.2m/s, Luhtanen, 1988; 17.6 years, 22.3-30.0 m/s, Rodano and Tavana, 1993; 16.8 years, 32.1m/s, Nunome et al., 2006a). Foot speeds in this study (21.3 m/s) were similar to the values reported by Nunome (2006a, 16.8 years, 22.7 m/s).

Junior kickers produced lower foot and ball speeds compared to senior players. These differences were substantial, with foot speed differing by 5.4 m/s (24% smaller) and ball speed by 7.9 m/s (32% smaller), and reflected by the effect size. Given Ball (2008b) reported foot speed was the most strongly correlated technical factor associated with distance ($r = 0.68$, $p < 0.01$), the difference in foot speed found in this study is likely to relate to a substantial difference in kick distance between juniors and seniors.

While foot and ball speeds were significantly different between junior and senior players, ball:foot ratio was not. This suggests that attempting to increase foot speed rather than focussing on the nature of impact is the more appropriate method of improving junior players. However, from a practical perspective, the difference in ratios represents a 6% difference in ball speed for the same foot speed. In terms of kick distance in the elite level, this is likely to represent a meaningful difference. Further, while junior and senior players produced significant correlations between foot and ball speed, these values were different for each group (senior $r = 0.79$ $r^2 = 62\%$, $p < 0.001$, junior $r = 0.57$, $r^2 = 32\%$, $p < 0.001$) and a substantial amount of variance was not accounted for in both groups. This can only be explained by impact factors such as ball orientation, position of ball on foot, behaviour of the foot due to impact with the ball and/or work done on the ball. As such, while foot speed should be a priority for development, impact factors should not be ignored.

Junior players performed less work on the ball than senior players. This was due to a combination of less average force applied to the ball (junior = 679 N; senior = 1023 N) and a smaller displacement over which force was applied to the ball (junior = 0.20 m; senior = 0.22 m). Work and foot speed were positively correlated (senior; $r = 0.84$, $r^2 = 71\%$, junior; $r = 0.55$, $r^2 = 30\%$) indicating that foot speed is an influencing factor in this generation of work. However, as for the relationship between foot and ball speed, there are clear differences between seniors and juniors.

Of interest from table 1 was that ball displacement was significantly smaller for the junior group while the change in shank angle was slightly larger (medium effect although not significant). This could be explained by one of three things: differences in knee linear velocity, differences in ankle plantar flexion, or differences in ball deformation at release. *Post-hoc* analysis of the kicks in this study showed that knee linear velocity did not differ and visual inspection of video indicated no difference in deformation of the ball at release. However, ankle plantar flexion was greater for junior players (Senior = 4°, Junior = 6°, $p < 0.003$). This might explain some of the variance unaccounted for in work on the ball by foot speed. Asami and Nolte (1983) and Sterzing and Hennig (2008) both reported that ankle plantar flexion occurred during ball impact in soccer and suggested a firmer foot (less flexion) is better for performance. This was questioned by Nunome et al (2006b) who reported observing a player that produced one of the highest ball velocities also produced one of the highest ranges of plantar flexion. However, given senior AF players plantar flexed less than juniors, a reduced plantar flexion would seem to be the better option in AF, supporting Asami and Nolte (1983) and Sterzing and Hennig (2008). This finding suggests that training to reduce plantar flexion during impact through kicking drills and conditioning of the ankle musculature might be beneficial to improving kick distance.

The underlying mechanisms for these impact differences between senior and junior players need exploration in future research. Attempting to link kinetic and kinematic factors with important impact factors is essential in developing useable coaching cues. Strength is another likely source of impact differences. This is another useful future direction, which should be explored through kinetic analysis of the kick itself, as recommended by Nunome et al. (2006a), as a more task specific measure of strength compared to isokinetic measures.

CONCLUSION: This study provides descriptive data for impact characteristics for junior AF players performing the drop punt. Junior kickers produced smaller foot and ball speeds as well as displacement of the ball while in contact with the foot and work on the ball compared to senior AF players. No statistical difference existed for ball to foot speed ratio although the differences are suggested to be practically significant. Juniors need to develop the ability to produce greater foot speeds and to apply greater force to the ball to develop their kicking to a senior level.

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ISBS 2010

Poster Session 1

INTRA-RATER AND INTER-RATER RELIABILITY OF A MODEL-BASED IMAGE-MATCHING MOTION ANALYSIS TECHNIQUE IN MEASURING ANKLE JOINT KINEMATICS

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The aim of this study was to assess the intra-rater and inter-rater reliability of the MBIM technique in measuring ankle joint kinematics. Three cadaveric below-hip specimens were prepared for performing full-range plantar/dorsiflexion, in/eversion and relative circular motion between the shank and foot segments. A detailed skeleton matching protocol was given to two researchers and each researcher performed the matching five times on each specimen. Intra-rater and inter-rater reliability were assessed with interclass correlation (ICC). The results showed excellent intra-rater reliability (ICC coefficient > 0.978) and excellent inter-rater reliability (ICC coefficient > 0.981). Therefore, the MBIM technique for analyzing ankle joint kinematics is repeatable and is a good motion analysis tool for sports science and sports medicine related research.

KEYWORDS: ankle joint movement, motion analysis technique

INTRODUCTION: Motion analysis is often used to measure the joint kinematics in sports science research. Motion analysis techniques must have good validity and reliability to ensure the data obtained are accurate and repeatable. Krosshaug and Bahr (2005) introduced a model-based image-matching (MBIM) technique for investigating human motion from uncalibrated video sequences and employed this technique to investigate knee joint kinematics in some sports injury incidents (Krosshaug et al. 2007). For measuring ankle joint kinematics, the validity of the MBIM technique has been demonstrated (Mok et al. 2009). However, since the MBIM technique requires manual skeleton matching, it depends on the researcher's ability to accurately match the orientation of each segment. As a result, measurement difference may exist within and between researchers. Before considering the MBIM technique as a reliable motion analysis tool, detailed matching instructions must be developed and its reliability must be assessed. Therefore, the aim of the present study was to assess the intra-rater and inter-rater reliability of the MBIM technique using a standard matching protocol proposed by our research team in measuring ankle joint kinematics.

METHOD: Three cadaveric below-hip specimens were prepared for testing. Each specimen was mounted on a jig in an upright position. Four high speed cameras (Casio EX-F1, Japan) were used to record the ankle motion in 30Hz with 640x480 resolutions from four different views. A static calibration trial in the anatomical position served as the offset position to determine the segment embedded axes of the shank and foot segment. The line connecting knee joint center and ankle joint center was the longitudinal axis of the shank segment (X1). The anterior-posterior axis of the shank segment (X2) was the cross product of X1 and the line joining the lateral femoral epicondyle and medial femoral epicondyle. The medial-lateral axis of the shank segment was the cross product of X1 and X2 (Wu et al. 2002).
Full-range plantar-flexion/dorsiflexion, inversion/eversion and shank circular motion were performed by moving the shank segment on the ankle joint manually. The ankle joint kinematics in each of the four views was analyzed using the MBIM technique (Krosshaug et al., 2005). Using a commercialized animation software Poser (Poser4, Curious Lab, US), a

virtual environment was built and matched with the video images in every camera view by adjusting the camera calibration parameters. A skeleton model (Zygote Media Group Inc, USA) was customized to match the anthropometry of the specimen. The skeleton model was matched frame by frame starting with the shank segment, followed distally by the foot and the toe segments. The joint angle time histories were read into Matlab (MathWorks, USA) with a customized script for data processing.
Two researchers, A and B, performed the manual skeleton matching process five times on each specimen. The researchers were with good human biomechanics knowledge and were trained to implement the skeleton matching with the same instructions (Table 1). In each frame, the skeleton model has to be matched such that it is in an anatomically accurate position and is contained within the image boundary (Figure 1). For plantar-flexion/dorsiflexion, inversion/eversion and internal/external rotation of the ankle joint, anatomical landmarks and joint orientations were used as indications of the direction of movement. Lastly, the motion of the skeleton model was reassessed for the whole video and adjusted frame by frame to ensure a smooth matched motion.

Table 1. Matching instructions given to researchers for the skeleton matching

Item		Instructions
1.	General	(a) Within image boundaries
		(b) Anatomically correct
		(c) Smooth motion
2.	Plantar flexion/dorsiflexion	(a) Identify the long axis of the shank segment
		(b) Identify the long axis of the foot segment
3.	Inversion/eversion	(a) Identify the plantar foot
		(b) Regard foot segment as a rectangular board
4.	Internal/external rotation	(a) Identify the patella position
		(b) Identify the anterior edge of shank

Figure 1. Matching of the skeleton model by the MBIM technique.

Intra-rater reliability was assessed by comparing the results of the five matching trials on each specimen performed by each researcher. Inter-rater reliability was assessed by comparing the first matching trial performed on the same specimen between the two researchers. A work distribution graph is shown on Figure 2. Intra-rater and inter-rater reliability were assessed with interclass correlation (ICC), two-way mixed measures assuming the interaction effect is absent (Hopkins, 2000).

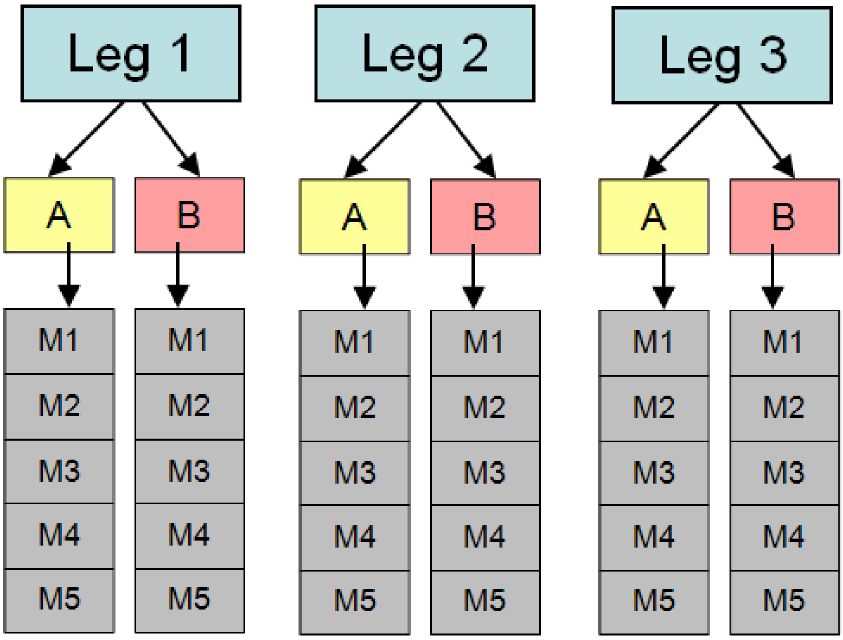

Figure 2. Work distribution of operator A and B on skeleton matching

RESULTS: The ICC coefficients for intra-rater reliability demonstrated excellent correlation (ICC coefficient > 0.978) between the kinematic data analyzed by the same operator (Table 2). For inter-rater reliability, the ICC coefficient also demonstrated excellent correlation (ICC coefficient > 0.981) between the kinematic data analyzed by the two operators (Table 3).

Table 2. Intraclass correlation for the intra-rater reliability of the kinematic data.

	Intraclass Correlation					
	Plantarflexion/dorsiflexion		Inversion/eversion		Internal/external rotation	
Researcher	A	B	A	B	A	B
Specimen 1	0.999	0.998	0.997	0.993	0.957	0.968
Specimen 2	0.997	0.999	0.999	0.999	0.991	0.987
Specimen 3	0.997	0.996	0.992	0.995	0.986	0.983
Average	0.998	0.998	0.996	0.996	0.978	0.979

Table 3. Intraclass correlation for the inter-rater reliability of the kinematic data.

	Intraclass Correlation		
	Plantarflexion/dorsiflexion	Inversion/eversion	Internal/external rotation
Specimen 1	0.987	0.972	0.956
Specimen 2	0.998	0.996	0.992
Specimen 3	0.996	0.994	0.995
Average	0.994	0.987	0.981

DISCUSSION: The aim of the present study was to assess the intra-rater and inter-rater reliability of the MBIM technique in measuring ankle joint kinematics. It is important to evaluate the repeatability of this technique since a manual skeleton matching process is required. Our research team has developed a standard protocol for analyzing ankle joint kinematics using the MBIM technique. Our protocol uses anatomical landmarks and joint orientations as indicators of joint movements to ensure matching accuracy.
Following our protocol, the average ICC coefficients for the intra-rater reliability were greater than 0.978 for all ranges of motion and the average ICC coefficients for the inter-rater

reliability were greater than 0.981 for three cadaveric specimens. These results imply that different trained researchers can produce the same results with excellent repeatability.
The MBIM technique is newly introduced and has the potential to be developed as a new on-field markerless motion analysis tool because calibrated capture volume and skin markers are not required. This technique can be used to analyze the kinematics of foot and ankle injuries from videos captured during televised sport events.

CONCLUSION: This study presented a model-based imaging-matching motion analysis technique in measuring ankle joint kinematics with excellent intra-rater and inter-rater reliability. This technique can reliably produce ankle joint kinematics profile from uncalibrated video sequences and can be used as an on-field markerless motion analysis tool to analyze real-game sporting motions.

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UNDERWATER NON-LINEAR CAMERA CALIBRATION: AN ACCURACY ANALYSIS

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KEYWORDS: non-linear camera calibration, 3D underwater analysis.

INTRODUCTION: One of the most challenging problems associated with underwater 3D movement analysis is the accurate calibration of the cameras. Additional sources of errors are present in underwater acquisitions such as the nonlinear distortion caused by water interface, camera lenses (ex. wide angle) and housing's glasses. Despite this, in the literature, systems based on a linear calibration model (DLT) were proposed (Yanai et al., 1996; Machtsiras & Sanders, 2009; Gourgoulis, et al. 2008). However, the results of underwater accuracy were not similar to those obtained out of the water. In Kwon, et al. 1999, the use of a modified DLT algorithm to model the distortion was proposed but the results of accuracy were not substantially improved, with Root Mean Square (RMS) values ranging from 5.6 to 7.2mm. Recently, alternative approaches were proposed to non-linear camera calibration and submillimeter accuracy was reached (Cerveri et al., 1998; Zhang, 2000; Pribanić, Sturm & Cifrek, 2008). However, these approaches were not applied underwater. In previous work, a new non-linear calibration method using a straight line plane object was proposed and tested out of the water (Silvatti et al., 2009 available in http://calib.googlecode.com). In this work, this novel method was tested in underwater conditions and its accuracy evaluated.

METHOD: A kinematic analysis system (DVideo, Figueroa et al., 2003) was adapted for underwater online data acquisition. The system consisted of a computer connected to two Basler cameras (50Hz, wide angle lens) enclosed in housings specially designed. Tripods were adapted with suction cups to fix it on the swimming pool floor. The cameras were synchronized by a gen-locked trigger. The non-linear camera calibration method required eight points with known coordinates to define the extrinsic parameters and the acquisition of the motion of a chess board to obtain the intrinsic and distortion parameters. The waterproof plane object (chess board) contained a 5x6 squares pattern defining straight orthogonal lines (100mm x 100mm with 42 corners). The chess board was moved in the whole acquisition volume ($4.5 \times 0.6 \times 1 m^3$) and tracked automatically in approximately 300 frames. The accuracy was evaluated in a dynamic test using a rigid bar and the follow variables were calculated: a) the mean absolute errors b) the standard deviation, c) the minimum and d) maximum error, e) the RMS of the distances between two markers (expected value=283.14mm) obtained as a function of time (15 seconds) and f) the RMS relative to reconstruction expressed as a percentage of the real length of the rigid bar's movement.

RESULTS: Table 1 shows the variable values: mean, standard deviation, minimum error, mean absolute errors, maximum error, RMS of the distance curves between markers and the %RMS for the proposed method in dynamic test.

Table 1. Results of the method proposed in the rigid bar dynamic test. Expected bar length 283.14mm. Values expressed in millimeter (mm).

	Mean	Standard Deviation	Minimum Error	Mean Absolute Error	Maximum Error	RMS	%RMS
Proposed Method	283.22	0.80	0.07	0.64	4.57	0.80	0.28

DISCUSSION: The mean absolute error presented in Yanai et al. (1996) was 5.12mm compared to 0.64mm in the proposed method. The %RMS found in Gourgoulis et al. (2008) was 1.28%, in a 4.5x2x1 m^3 acquisition volume, and 0.65% in a 1x1x1 m^3 acquisition volume. The proposed method reached a better value in both cases (0.28%) with an acquisition volume of 4.5x0.6x1m^3. Using the regular DLT method, Kwon et al. (2000) obtained RMS and maximum error values of 39.3mm and 17.4mm, respectively. Applying a new modified DLT algorithm including distortion correction parameters and the RMS value ranged from 5.6mm to 7.2mm and maximum error ranged from 9.3mm to 9.7mm, compared to the 0.8mm and 4.57mm in the proposed method. The results of the proposed method can be compared to high accurate calibration method in out of the water condition. According to Pribanić et al., (2008) the accuracy found in two non-linear camera calibration methods (wand calibration and 2D plane model) were smaller than 1mm. Another advantage in the proposed method is the simplicity and the portability of the system due to the requirement in terms of calibration objects. The proposed method required only a waterproof planar chess pattern, previously assembled, and few markers with known coordinates. Most calibration methods used in Biomechanics require the construction and transportation of rigid volumetric structures with many markers. This kind of object is very difficult to measure when large volumes are involved.

CONCLUSION: The results of the proposed method provided better underwater accuracy compared to all previous papers reported in the literature. The results of the underwater chess method revealed to be an applicable alternative with good accuracy and portability for underwater non-linear camera calibration.

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GROUND REACTION FORCE AND RATE OF FORCE DEVELOPMENT DURING LOWER BODY RESISTANCE TRAINING EXERCISES

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This study quantified the differences in the kinetic profiles of the back squat, deadlift, step-up, and lunge. Eleven subjects performed 2 repetitions of their 5 repetition maximum in each of the 4 exercises. Kinetic data were collected using a force platform. The exercises were compared based on their peak vertical ground reaction force (GRF_P) and rate of force development (RFD) in both the eccentric and concentric phase. A repeated measures ANOVA indicated differences ($p \leq 0.001$) in GRF_P attained for the different exercises in both the eccentric and concentric phase. No significant differences ($p \geq 0.05$) were found for RFD for any of the exercises in either the eccentric or concentric phase. Results can guide the development of training programs that are specific to strength, explosiveness, or osteogenesis.

KEYWORDS: kinetics, strength, explosiveness, osteogenesis

INTRODUCTION: Understanding the relative value of an exercise is important in the design of a training program for strength, explosiveness, or osteogenesis. In order to perform any exercise, a certain level of force must be applied. Therefore, ground reaction force (GRF) and its derivatives, peak GRF and RFD, have been used to compare exercise characteristics (Ebben et. al, 2009; Escamilla et. al, 2002; Jensen and Ebben, 2002; Salem et. al, 2004; Wilson et. al, 2008; Zink et. al, 2006).

Previous research has used GRF to analyze performance of single or variations of single resistance exercises such as the squat (Zink et. al, 2006), deadlift (Escamilla et. al, 2002), step-up (Salem et. al, 2004), and lunge (Wilson et. al, 2008).

Research has also evaluated two or more resistance training exercises in a single study. Kinetic analysis has been used to compare the hang clean and hang snatch (Jensen and Ebben, 2002). Multiple exercises have also been studied to determine training load predictions (Ebben et. al, 2008) and muscle activation during multiple lower body resistance exercises (Ebben, 2009). Research has also analyzed the GRF of multiple modes of exercise such as walking, running, plyometrics, and the back squat (Ebben et. al, 2009).

Resistance training is used for many purposes such as performance enhancement or rehabilitation. Peak GRF and rate of force development (RFD) can be used as an indicator of the exercise's potential to increase strength and explosiveness, respectively. A secondary effect of resistance training is the bone adaptation that results from high strains and magnitudes of resistance. While no research has produced precise prescription for osteogenesis, it has been proposed that higher magnitude and rate of loading produce greater adaptation (Skerry, 1997). Exercise GRF and RFD can be used to estimate the magnitude and rate of loading, respectively, of resistance training exercises (Ebben, et al, 2010).

No study has compared kinetic aspects of multiple resistance training exercises. Therefore, the purpose of this study is to compare the kinetic characteristics of the squat, deadlift, step-up, and lunge. Peak vertical GRF (GRF_P) and RFD during the eccentric and concentric phase of each exercise will be assessed. These results will help determine optimal exercises to be included in training focused on strength, explosiveness, or osteogenic potential.

METHODS: Eleven recreationally active subjects participated in this study (mean ± SD; age

= 22.45 ± 2.70 years; body mass = 83.12 ± 15.19 kg). Subjects included males who were 18-45 years old and participated in at least 6 weeks of lower body resistance training prior to testing. All subjects provided written informed consent and the study was approved by the university's internal review board.

A pre-test habituation session including assessment of 5 repetition maximum (RM) loads for back squat, deadlift, step-up, and lunge. After at least 48 hours of recovery, subjects returned for the test session and performed 2 repetitions of the subject's 5RM load in each of the randomly ordered test exercises with 5 minutes of rest between each set. Back squats were performed to the depth of a measured 90 degree knee angle. Step-ups were standardized to an 18 inch box. Lunges were performed to a length of 120 cm. All step-up and lunge sets were performed with only the right leg. The deadlift and step-up were performed with the eccentric muscle action first followed by the concentric muscle action. This was done in order to keep consistency in the order of muscle actions of all exercises being studied. Instructions were given to perform exercises at maximal velocity to enhance external validity.

Prior to the pre-test habituation and test sessions, a warm-up was performed consisting of 3 minutes on a cycle ergometer proceeded by dynamic stretching exercises including 5 repetitions of each of the following: slow and fast body weight squats, forward and backward lunges, and walking quadriceps and hamstring stretches. A 20 yard skip and 5 sets of 10 yard sprints gradually increasing in speed were also performed. Warm-up sets of 5 reps at 60% and 3 reps at 80% of a self-predicted 1 RM were completed for the first exercise to be performed. Warm-up sets were not performed for the rest of the exercises due to already being acclimated to high intensity loads and to minimize fatigue.

The test exercise modes were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA, USA) that was bolted to the laboratory floor and mounted flush in the center of a 122 x 244 cm weightlifting platform. The force platform was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA USA) for later analysis. The GRF_P and RFD were calculated from the force-time records for both eccentric and concentric phases. All values were determined as the average of two repetitions for each exercise. The GRF_P was defined as the highest value attained. The RFD was defined as GRF_P minus the GRF 250 ms prior to the peak divided by the time elapsed between these two values (250 ms) and calculated consistent with methods previously used (Jensen et al, 2008; Ebben et al, 2010).

All data were analyzed with SPSS © (Version 16.0). A repeated measures ANOVA was used to determine possible differences in GRF_P and RFD during the eccentric and concentric phase between the exercise modes, as well as differences in estimated 1 RM. Significant main effects were further analyzed with Bonferroni adjusted pairwise comparison to identify the specific differences between the exercise modes. A Pearson's correlation coefficient was used to assess the relationship between the exercise load and the GRF_P and RFD. The *a priori* alpha level was set at $p \leq 0.05$. Power and effect sizes are listed as d and partial eta squared (η_p^2), respectively.

RESULTS: Analysis of GRF_P showed significant main effects for the eccentric ($p \leq 0.001$, d = 1.00, η_p^2 = 0.73) and concentric ($p \leq 0.001$, d = 1.00, η_p^2 = 0.91) phases, indicating differences in force requirements between the exercises. No significant main effects were found for the RFD during the eccentric (p = 0.78) or concentric (p = 0.51) phases. Significant main effects were also found ($p \leq 0.001$, d = 1.00, η_p^2 = 0.86) for estimated 1 RM. Post hoc analysis identified the specific differences in GRF_P and estimated 1RM between the exercises (Table 1 and 2). Descriptive RFD information is also presented in Table 3 and 4. Squat, deadlift and lunge eccentric and concentric GRF_P were correlated with exercise load ($p \leq 0.01$, R ranged from 0.76 to 0.93). Step-up eccentric and concentric GRF_P and RFD for all exercises were not correlated to exercise load.

Table 1. Eccentric and Concentric GRF_P in Newtons (N) (mean ± SD). N=11.

	Squat	Deadlift	Lunge	Step-Up
ECC	2440.19 ± 293.70*	2301.41 ± 371.18*	1714.87 ± 351.87**	1575.08 ± 490.75**
CON	2646.44 ± 391.16***	2539.07 ± 458.90***	1910.37 ± 428.82*	1587.34 ± 372.29*

* = Different than all other exercises; $p \leq 0.05$
** = Different than Squat and Deadlift; $p \leq 0.05$
*** = Different than Lunge and Step-Up; $p \leq 0.05$

Table 2. Estimated 1RM in kilograms (kg) (mean ± SD). N=11.

Squat	Deadlift	Lunge	Step-Up
169.99 ± 25.94*	163.00 ± 30.49*	76.99 ± 38.41**	75.13 ± 22.10**

* = Different than Lunge and Step-Up; $p \leq 0.001$
** = Different than Squat and Deadlift; $p \leq 0.001$

Table 3. Eccentric rate of force development ($N \cdot s^{-1}$) (mean ± SD). N=11.

Step-Up	Squat	Lunge	Deadlift
2312.96 ± 4857.66	2095.73 ± 3052.86	2022.97 ± 1240.32	1172.98 ± 363.27

Table 4. Concentric rate of force development ($N \cdot s^{-1}$) (mean ± SD). N=11.

Deadlift	Step-Up	Lunge	Squat
2373.98 ± 1900.19	2162.10 ± 2389.72	1950.02 ± 1923.76	1431.39 ± 723.24

DISCUSSION: This is the first study to kinetically quantify and compare multiple resistance training exercises, including the back squat, deadlift, step-up, and lunge. This study shows significant differences in GRF_P between exercises in both the eccentric and concentric phase. The concentric phase GRF_P was highest in the back squat followed by the deadlift. The lunge and step-up were significantly lower than the back squat and deadlift, but no differences existed between them. Significant differences were not found for either the eccentric or concentric RFD, although substantial mean differences were present.

Differences in GRF_P may be explained, in part, by the load used for each exercise. In this study, the squat and deadlift mean 1 RM were significantly higher than the lunge and step-up. In fact squat, deadlift, and lunge exercise load was significantly correlated with GRF_P.

It is noted that the exercises with higher GRF_P such as the squat and deadlift, are performed with bilateral weight distribution. Exercises with lower GRF_P such as the lunge and step-up are performed unilaterally. The unilateral component creates a smaller base of support and reduces maximal loading, consistent with previous research (McBride et. al; 2006), and ultimately reducing GRF_P in the present study.

Statistically significant differences did not occur in the RFD between the 4 exercises, although substantial mean differences exist. For example, the deadlift produced a 65.8% higher concentric RFD than the back squat. The step-up and lunge also have a 51.0% and 36.3% higher mean RFD than the back squat, respectively, potentially due to the unilateral nature and the bilateral deficit phenomenon (Hay et. al, 2006).

While the back squat produces the greatest concentric GRF_P it also has the slowest mean concentric RFD. The relationship between GRF_P to RFD is consistent with the force velocity curve, which estimates that the greater the load lifted, the slower the movement of the exercise (Kraemer & Spiering, 2006). The greater motor unit recruitment resulting from slower movements results in maximal strength development (Campos et. al, 2002). Therefore, strength focused training should prioritize the back squat over exercises such as the lunge and step-up.

Training exercises should also simulate the sport being trained for. Sport-specific simulation includes the relative speed of the exercise for muscle adaptation (Kraemer & Spiering, 2006) as well as similar movements for neural adaptation (Behm, 1995). While the step-up and lunge do not allow for the use of comparatively high loads, they may still be valuable due to a high RFD and their unilateral nature. Unilateral exercises such as steps have relatively high mean concentric RFD and may be useful for training athletic activities such as sprinting. Training for powerful bilateral movements such as vertical jumping in basketball or volleyball should incorporate deadlifts for a bilateral high concentric RFD exercise.

The results of this study provide some insight into the potential of osteogenesis from these exercises. The squat's high GRF_P estimates a high magnitude of load, and the deadlift's high concentric RFD approximates a greater rate of loading, both of which are believed to be important for osteogenesis (Skerry, 1997). Therefore, a combination of these exercises should be included in programs designed to promote osteogenesis

CONCLUSION: This study shows that the back squat has the highest mean GRF_P followed by the deadlift, lunge, and step-up. Mean eccentric RFD is highest in the step-up followed by the squat, lunge, and deadlift. Mean concentric RFD is highest in the deadlift followed by the step-up lunge and squat. High GRF_P is necessary for strength training while high RFD is essential for explosive training. Many sports will require both strength and explosiveness.

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DYNAMIC STABILIZATION DURING THE LANDING PHASE OF PLYOMETRIC EXERCISES

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This study examined the differences in and the reliability of time to stabilization (TTS) of several plyometric exercises. Twenty six men performed a variety of plyometric exercises representing a continuum of intensities of landing instability, including line hops, cone hops, squat jumps, tuck jumps, countermovement jumps, dumbbell countermovement jumps, and single leg countermovement jumps on a force platform. A repeated measures ANOVA with Bonferroni post hoc corrections was used to evaluate the differences in TTS between plyometric exercises. Practitioners who use plyometrics to train dynamic stability and balance should create programs that progress the intensity of the exercises based on the results of this study. This study also demonstrates that TTS reliability is fair to excellent for a variety of jumping conditions.

KEY WORDS: stretch shortening cycle, reliability, balance, postural control

INTRODUCTION: Balance training improves postural stability and may reduce injury (Wikstrom 2009). Plyometric training offers promise as a balance training stimulus, though many aspects of plyometric program design are unclear. Plyometric intensity is among the most important variables for designing a program (Ebben et al., 2008; Jensen & Ebben, 2007). Studies have assessed intensity of plyometric exercises using electromyographic, kinematic, or kinetic analysis (Ebben et al., 2008; Jensen & Ebben, 2007). However, previous research assessing plyometric exercises did not assess their characteristics with respect to postural control or stability. The intensity of the landing phase of plyometric exercises may be measured by the difficulty of dynamic postural stabilization and quantified by time to stabilization (TTS).

Time to stabilization is derived through force plate data and used to evaluate postural stability by measuring the time taken for vertical ground reaction force to reach and stabilize within 5% of the subject’s body weight following the landing from a jump (Wikstrom, 2004). Time to stabilization has been used to examine stability during ankle taping and bracing (Jacobs et al., 2006) and to compare the effect of different levels of functional ankle instability (Wikstrom et al., 2004; Ross et al., 2005). Time to stabilization has typically been used with a jump landing protocol that included a bilateral take off from 70 cm away from a forceplate, jumping to approximately 50% of the subjects maximal vertical jump, and landing on a single leg while attempting to stabilize as quickly as possible (Wikstrom et al, 2004; Wikstrom et al., 2005; Ross et al., 2005). Functional ankle instability has also been assessed using TTS after subjects stepped down from a 20 cm box (Wikstrom et al., 2005). Others have used TTS to characterize the differences between dynamic stability of athletes from different sports and to assess gender differences in this variable (Butcher-Mokha et al., 2005). Few studies assessed the reliability of TTS. Flanagan et al. (2008) demonstrated average measures intraclass correlation coefficient values of 0.68 for TTS during the landing from a 30 cm depth jump.

The purpose of this study was to quantify the postural stability demands of the landing phase of a variety of plyometric exercises by assessing vertical TTS. This study also sought to further identify plyometric exercise characteristics with TTS and to assess the reliability of this variable during a variety of jump landing conditions.

METHODS: Twenty four athletic or recreationally active adults (mean ± SD; age = 20.23 ± 1.63 years, height = 180.98 ± 6.13 cm; body mass = 79.41 ± 9.03 kg; vertical jump = 63.55 ± 5.80 cm; back squat 5 RM 131.37 ± 25.64 kg) volunteered to serve as subjects for the study. The subjects were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board.
All subjects performed a habituation and testing session. Prior to each session, the subject warmed up with 3 minutes of low intensity work on a cycle ergometer and performed dynamic stretching exercises. During the habituation session, subjects performed 2 repetitions of the countermovement jump which was assessed using a Vertec (Sports Imports, Columbus, OH, USA). Subjects then rested for 4 minutes and performed their 5 repetition maximum (RM) back squat test in order to obtain a strength measure that further characterizes the subjects training status. Subjects were instructed and provided a demonstration on the correct performance of the plyometric exercises to be assessed during the test session. Subjects then performed each of these exercises until they demonstrated the correct performance of the technique. The plyometric exercises included the lateral line hop (LH), 15.24 cm lateral cone hop (CH), squat jump (SJ), tuck jump (TJ), countermovement jump (CMJ), loaded countermovement jump with dumbbells equal to 30% of the subjects previously assessed estimated 1 RM squat (DBJ), and the single leg jump (SLJ). After the habituation session, subjects recovered for at least 48 hours and returned for the test session.
During the testing session, subjects warmed up with the same protocol they used prior to the habituation session. Subjects performed 3 repetitions of each of the test plyometric exercises in a randomized order with 1 minute of rest between each exercise, which insures full recovery (Read & Cisar, 2001). Subjects were instructed to stabilize their landing as quickly as possible, with knee and hip flexion, face straight ahead, remain motionless for a period of 5 seconds, and limit upper limb movement upon landing consistent with previous recommendations (Flanagan et al., 2008).
The test exercises were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies, Inc., Watertown, MA, USA), which was calibrated with known loads to the voltage recorded prior to the testing session. Data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA, USA) for later analysis. All values were determined as the average of three repetitions for each of the plyometric exercises. Vertical TTS was determined consistent with methods previously used (Flanagan et al., 2008; Wikstrom et al., 2005) as demonstrated in Figure 1.

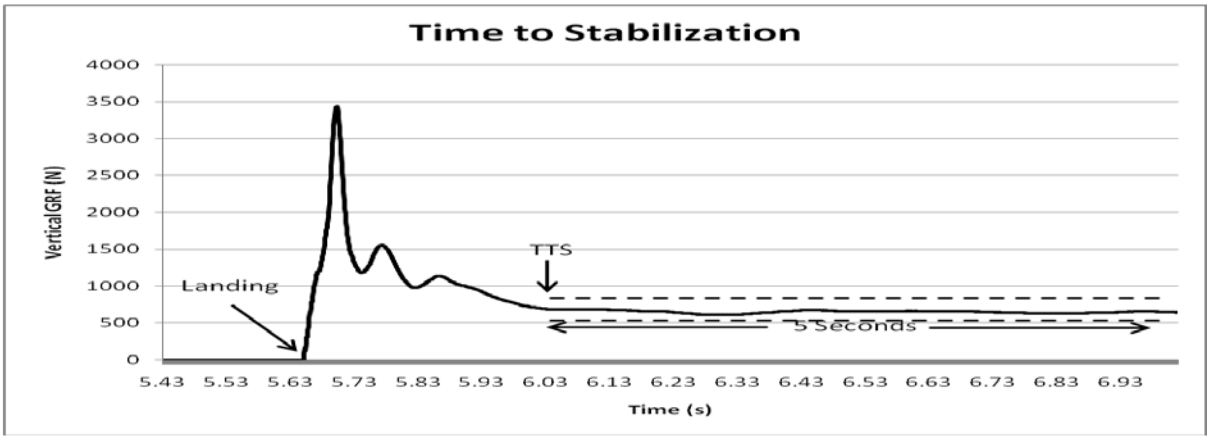

Figure1. Sample force time record of time to stabilization

A repeated measures ANOVA with Bonferroni adjusted pairwise comparison was used to evaluate differences in TTS of the plyometric exercises. The trial to trial reliability of TTS was assessed for each plyometric exercise with the intraclass correlation coefficient (ICC). In addition, repeated measures ANOVA was used to confirm that there was no significant difference (P >0.05) between the three trials. The statistical analyses were undertaken with SPSS 17.0. Assumptions for linearity of statistics were tested and met. An *a priori* alpha level of $p \leq 0.05$ was used with effect size and power represented by η_p^2 and d, respectively.

RESULTS: The analysis of TTS revealed significant main effects for test condition ($P \leq 0.001$, $\eta_p^2 = 0.41$, $d = 1.00$) and specific differences in TTS between a number of plyometric exercises (Table 1). Results of the reliability analysis show ICC ranging from 0.51 to 0.86 with no significant differences between the trials (P > 0.05).

Table 1. Time to stabilization expressed in seconds of men subjects during a variety of plyometric exercises (N=26)

LH	CH	SJ	DBJ	CMJ	TJ	SLJ
0.50[a] ± 0.09	0.57[a] ± 0.15	0.67[b] ± 0.14	0.73[c] ± 0.18	0.73[d] ± 0.21	0.77[b] ± 0.16	0.99[a] ± 0.39

LH = line hop, CH = 15.24 cm cone hop, SJ = squat jump, DBJ = dumbbell countermovement jump, CMJ = countermovement jump, TJ = tuck jump, SLJ = single leg jump
[a] Significantly different than all other plyometrics
[b] Significantly different than all other plyometrics except for DBJ and CMJ.
[c] Significantly different than all other plyometrics except for SJ, CMJ, TJ
[d] Significantly different than all other plyometrics except for SJ, DBJ, TJ

DISCUSSION: This is the first study to assess the TTS of a variety of plyometric exercises demonstrating a number of differences with respect to this variable. Practitioners can use this knowledge to create performance enhancement and rehabilitation programs that progress instability, and thus the intensity of plyometric exercises. This study is also the first to demonstrate that TTS is fair to good, to excellent, based on the classifications of Fleiss (1986) (less than 0.4 was poor, between 0.4 and 0.75 was fair to good, and greater than 0.75 was excellent).

The SLJ produced the highest TTS of the plyometric exercises assessed. This finding is likely due to the smaller base of support associated with SLJ landing since the size of the base of support has been thought to affect postural control (Wikstrom et al., 2006). Previous research has demonstrated SLJ ground reaction forces and knee joint reaction force values that were considerably more than half of the values demonstrated for all other plyometric exercises performed in a bilateral condition (Jensen & Ebben, 2007). These findings confirm that the SLJ is the most intense plyometric exercise, and should be prescribed later in plyometric programs that seek to progress the exercise intensity.

Tuck jumps produced relatively high TTS demonstrating that this is a high intensity plyometric exercise, consistent with previous research assessing peak ground and knee joint reaction forces (Jensen & Ebben, 2007). This finding shows that not all “jumps in place” are low intensity plyometric exercises as previously suggested (Potach & Chu, 2008). The TJ requires hip flexion during ascent and consequently considerable hip extension during the descent phase in order to reposition the legs for landing. This action decreases the time to prepare for landing and likely increases the TTS. Pike jumps are likely to be similar since this exercise and the TJ share similar characteristics of active hip flexion followed by extension and a reduction in landing preparation and has been shown to be similar (Jensen & Ebben, 2007). Dumbbell countermovement jumps demonstrate moderate mean TTS values compared to the other exercises potentially due to the dumbbells decreasing extraneous hand movements which has been thought to increase TTS (Flanagan et al., 2008).

The SJ mean TTS was only higher than LH and CH, potentially due to the fact that the SJ does not activate the stretch-shortening cycle resulting in lower jump heights and consequently lower TTS compared to most exercises which may be why LH and CH TTS values are low.

The specific TTS values attained for the plyometric exercises in the present study (0.52 to 0.88 seconds) were shorter than other studies where values ranged from 0.65 to 2.7 seconds (Butcher et al., 2006; Flanagan et al., 2008; Jacobs et al., 2006; Ross et al., 2005; Wikstrom et al., 2004). Shorter mean vertical TTS in the present study may be a function of the subject training status, habituation, and differences in landing tasks between the studies. Reliability of the exercises in the present study may also have been affected by the requirement that subjects skillfully and reliably perform 7 different plyometric exercises, unlike previous studies where subjects performed only one or two (Butcher et al, 2006; Flanagan et al., 2008; Jacobs et al., 2006; Ross et al., 2005; Wikstrom et al., 2004).
Only one known study examined the reliability of vertical TTS (Flanagan et al., 2008). Finding high levels of reliability for repeated trials of dynamic postural stability measures of jump landings is inherently difficult. Any outcome measure that is dependent on proprioceptive and kinesthetic feedback, as well as reflexive and voluntary muscle responses (Wikstron et al., 2006) is likely to be rife with variability. On the other hand, any outcome measure that reduces the integrative sensorimotor complexity may be more internally controlled and potentially more reproducible, but at the cost of external validity. Ground reaction force derived measures of dynamic stability remain more reliable than other options including stabilometery techniques such as center of pressure measurements (Wikstrom, 2006). In the present study, TTS was consistently, moderately reliable, and in some cases, highly reliable, while providing strong external validity.

CONCLUSION: Practitioners should create plyometric programs that progress the intensity of the stability stimulus based on the results of this study. The plyometric exercises that produce the shortest TTS should be prescribed early in the program since these exercises provide the lowest challenge to dynamic stability. Plyometric exercises with increasingly higher TTS should be added as the program progresses. This study also demonstrates that TTS is moderately to highly reliable for a variety of jumping conditions.

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QUANTIFYING sEMG IN PRE-FATIGUE AND FATIGUE STATES DURING THE FASTBALL BASEBALL PITCH

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Injuries to baseball pitchers typically occur as a result of constant repetitive overuse. In attempt to eventually prevent overhead throwing overuse injuries, it is important that the biomechanics and muscle activations are understood. Therefore, the purpose of this study was to describe upper and lower extremity muscle activations involved in the pitching motion while in a pre-fatigue and fatigued state. Fourteen male pitchers volunteered to participate in the study. Participants were analyzed with surface electromyography and motion analysis software. The muscle firing patterns were described during the phases of the baseball pitch while in a pre-fatigued and fatigued state.

KEYWORDS: electromyography, overhand pitching, muscle activation

INTRODUCTION: In attempt to understand baseball pitching, it is important to understand the muscle activations that allow for proper pitching mechanics. Thus the understanding of primary muscle activations involved in the pitching motion is imperative to any type of injury prevention. The repetitive nature of the baseball pitch creates great challenges for sports medicine clinicians in an attempt to reduce the incidence of injury and in particular reduce the rate of overuse injuries (Fleisig et al., 1995; Lyman et al., 2002; Fortenbaugh et al., 2009). It is not only important to understand muscle activations during the pitching motion, but it would be beneficial to understand the muscle activations during a fatigued state. It has been discussed that fatigue to the scapular stabilizers, as a result of a repetitive pitching performance, can contribute to pathomechanics of the entire glenohumeral joint (Limpisvasti et al. 2007). Improving the understanding of the pitching motion will eventually allow for measures of injury prevention to be implemented. Fatigue or overuse often leads to injury in pitchers (Mullaney et al., 2005); however, there are no data examining muscle activations while fatigued during a pitching performance. Therefore, it was the purpose of this study to examine both upper and lower extremity muscle activations during baseball pitching while in the pre-fatigue and fatigued states.

METHODS: Fourteen baseball pitchers (16.0 ± 2.28 years, 178.2 ± 8.7 cm and 78.9 ± 17.55 kg) volunteered to participate in the current study. All participants had recently finished their competitive spring high school baseball seasons, and were deemed appropriately conditioned for data collection. Throwing arm dominance was not a factor contributing to participant selection or exclusion. All data collection sessions were conducted indoors at the University's Health, Physical Education, and Recreation building and were designed to best simulate a competitive setting. All testing protocols used in the current study were approved by the University's Review Board.
Location of the bilateral gluteus maximus, bilateral gluteus medius, throwing arm biceps, triceps, deltoid and scapular stabilizers were identified through palpation. Prior to testing, the identified locations for surface electrode placement were shaved, abraded and cleaned using standard medical alcohol swabs. Subsequent to surface preparation, adhesive 3M Red-Dot bipolar surface electrodes (3M, St. Paul, MN) were attached over the muscle bellies and positioned parallel to muscle fibers using techniques described by Basmajian and Deluca (1985). Following electrode placement, manual muscle tests (MMT) were conducted using techniques described

by Kendall et al. (1993). Manual muscle tests were used to identify the participant's maximum voluntary isometric contraction (MVIC) to which all sEMG data were compared.
Surface electromyographic (sEMG) data were transmitted to The MotionMonitor™ motion capture system (Innovative Sports Training Inc, Chicago IL) via a Noraxon Myopac 1400L 8-channel amplifier. The signal was full wave rectified and smoothed based on the smoothing algorithms of root mean squared at windows of 100 ms. Throughout all testing, sEMG data were sampled at a rate 1000 Hz. All sEMG data were notch filtered at frequencies of 59.5 Hz and 60.5 Hz respectively (Blackburn and Pauda, 2009).
In addition to sEMG data, kinematic data were collected to event mark the phases of the pitching motion. Kinematic data were collected using The MotionMonitor™ motion capture system (Innovative Sports Training, Chicago IL). Participants had ten electromagnetic sensors attached at the following locations: (1) the medial aspect of the torso at C7; (2) medial aspect of the pelvis at S1; (3) the distal/posterior aspect of the throwing humerus; (4) the distal/posterior aspect of the throwing forearm; (5) the distal/posterior aspect of the non-throwing humerus; (6) the distal/posterior aspect of the non-throwing forearm; (7) distal/posterior aspect of stride lower leg; (8) distal/posterior aspect of the upper stride leg; (9) distal/posterior aspect of non stride lower leg; and (10) distal/posterior aspect of non stride upper leg (Myers et al., 2005). Following the attachment of the electromagnetic sensors, an eleventh sensor was attached to a wooden stylus and used to digitize the palpated positions of the bony landmarks.
Participants were allotted an unlimited time to perform their own specified pre-competition warm-up routine. Participants were asked to spend a small portion of their warm-up throwing from the indoor pitching mound to be used during the test trials. After completing their warm-up and gaining familiarity with the pitching surface, each participant threw a series of maximal effort fastballs for strikes toward a catcher located the regulation distance (18.44 m). The mound was positioned so that the participant's stride foot would land on top of the 40 x 60 cm Bertec force plate (Bertec Corp, Columbus, Ohio) which was anchored into the floor. After five fastballs for strikes were thrown, the participants then threw a 2kg ball into a rebounder until they reported max perceived fatigue. A scale of 0-3 (Kimura et al., 2007), with three being only able to make 15 more throws, was used to quantify fatigue. Once a fatigue of 3 was reported, participants completed 10 more throws with the 2kg ball before returning to the mound to throw five maximum effort fastballs while in the fatigued state. Those data from the fastest pitch passing through the strike-zone for the pre-fatigue and fatigue deliveries were selected for analysis.
Data were analyzed in the current study using the statistical analysis package SPSS 15.0 for Windows. Data for the fastest strike mean and standard deviation for all sEMG and kinematic parameters were calculated for both pre-fatigue and post fatigue states. Once measures of central tendency were calculated, a series of descriptive statistics were conducted. A MANOVA was run for each of the four phases comparing pre-fatigue to fatigue with an adjusted alpha level using Bonferroni correction to allow for multiple tests ($p \leq 0.01$).

RESULTS: The pitching motion has previously been described into five phases (DiGiavine et al, 1992), however for the case of this study only stride through deceleration phases were analyzed. Stride phase was described as the beginning of motion to stride foot contact (SFC). The cocking phase was from SFC to maximum external rotation (MER) of the throwing shoulder. Next, was the acceleration phase that was from MER to ball release (BR), and finally phase 4, the deceleration phase, was described from BR to maximal internal rotation (MIR) of the throwing shoulder. There were no significant differences between pre-fatigue and fatigued states by phases ($p \leq 0.01$). Means of muscle activations are graphically summarized in Figures 1-2.

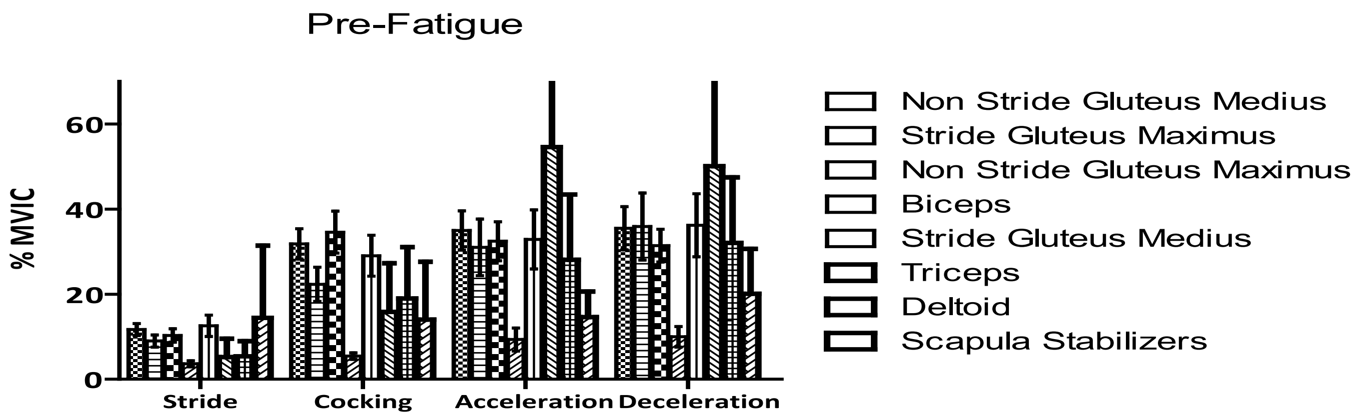

Figure 1. Muscle Activation as a %MVIC during the pre-fatigued state of the Baseball Pitch

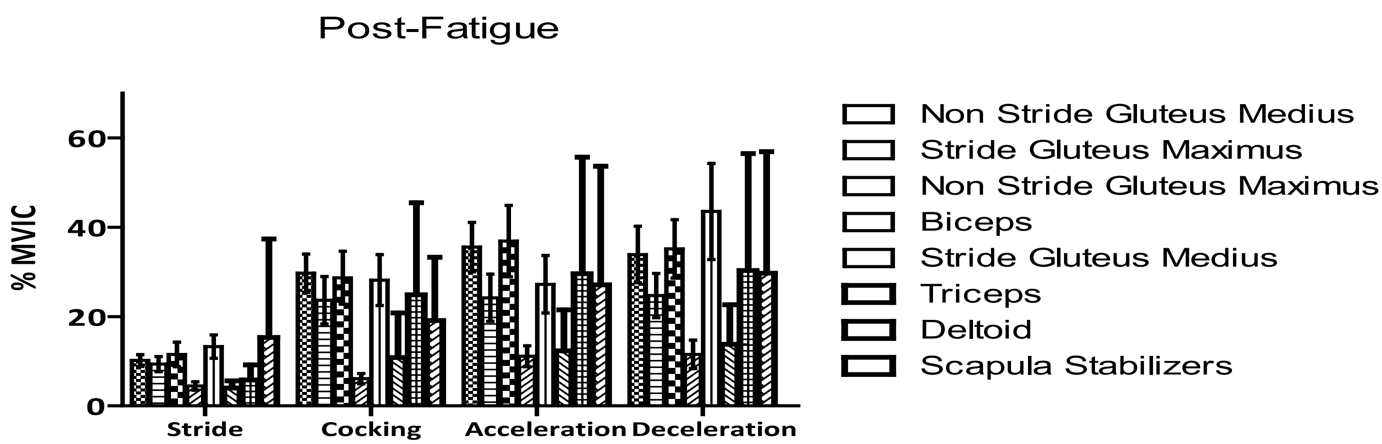

Figure 2. Muscle Activation as a %MVIC during the fatigued state of the Baseball Pitch

DISCUSSION: Previous sEMG studies of baseball pitching have focused on the upper extremity in a non-fatigued state. The current study was able to quantify muscle activations, of both the upper and lower extremities, while the participants were deemed free from fatigue and then in a fatigued state while throwing the fastball baseball pitch.

During pre-fatigue the stride leg and non stride leg had similar gluteal activations during pre-fatigue in attempt to stabilize the pelvis during the stride phase. The non stride gluteal group demonstrated greater activation as the participants were preparing for single leg support during pre-fatigue. Once fatigued, the stride leg gluteus medius had the greatest activation of the lower extremity muscles.

Pre-fatigue the gluteal muscle group increased in activation as did upper extremity muscle activations during cocking. The triceps displayed the greatest activation, eccentric in nature, during the cocking phase. Once fatigued, the triceps greatly reduced their activation; while the biceps, deltoid, and scapula stabilizers increased in activation. During the cocking phase the humerus was abducted and scapular movement during this phase allows for elevation of the acromion.

Pre-fatigued displayed a continual increase in gluteal muscle activation with the highest muscle activation being generated by the scapular stabilizers during acceleration. During cocking and acceleration the scapula must rotate in order for the rotator cuff to clear the acromion. In attempt to gain maximum external rotation, the scapula must retract and then protract to achieve acceleration for ball release. Once fatigued, the stride gluteus medius remained consistent in attempt to counter balance the non stride gluteal activation as well as the decreases activation of the stride leg gluteus maximus in attempt to keep the pelvis level while the body was shifting weight over to the stride leg for BR. Additionally, the deltoid and scapula stabilizers remained active.

Deceleration, pre-fatigue, revealed great activation of the gluteal muscle group with decreases in stride leg gluteal muscle activation during the fatigued state. The scapula stabilizers continued to stay active with slight decreases during fatigue. The scapula is the common point of attachment for the biceps, triceps, and deltoid. Therefore, the scapula stabilizers direct the efficiency of the musculature attached to the scapula. If the scapula is unstable, then alterations in all musculature that attaches to the scapula will occur.

CONCLUSIONS: We were able to identify muscle activation for the upper and lower extremities during the baseball pitch. We were also able to identify muscle activation during a fatigued state. Those muscles that had increased activation during the fatigued state could be a result of having to recruit more muscle fibers to perform the precision task of throwing a strike. To date there are no data available examining sEMG of the upper and lower extremity during the pitching motion in a pre-fatigued and fatigued state. More studies need to be conducted in attempt to validate our results with higher level of evidence of muscle activations during dynamic muscle fatigue. Studies examining kinematics and sEMG during the fatigue state are warranted. Findings from these types of fatiguing studies could direct practitioners in appropriate strength and conditioning regimens that would aid in pitching performance.

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KINEMATIC ANALYSIS OF LOWER LIMB IN FUTSAL BALL KICKING

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KEYWORDS: Futsal, kicking motion, image analysis

INTRODUCTION: The diameter of the futsal ball (200 mm) is smaller than that of the soccer ball by 20 mm, and the futsal ball also has lower resilience than the soccer ball. Because of these differences in the balls, it is thought that the kicking motions of futsal players are distinct from those of soccer players. No study has yet been conducted on the motion involved in kicking a futsal ball. The aim of this study was to clarify the difference between the motion involved in kicking the futsal ball with that involved in kicking the soccer ball.

MERHODS: The study population comprised 9 male professional futsal players who were instructed to kick a futsal ball and a soccer ball with maximum effort, using instep kicks. All the subjects kicked the balls at least 5 times, and the shot that involved maximal initial velocity was selected for further analysis. Three synchronized high-speed cameras (250fps) were used to record lower limb motion during the kicking. A digitizing system was used to manually digitize anatomical landmarks and the balls on the images recorded by these 3 cameras. Using these digitized data, we calculated the angle of the pelvis (defined as the vector from the left hip to the right hip and relative to the anterior direction within the horizontal plane), angle of the thigh-shank plane (defined as the vector normal to and pointing outward from the thigh-shank plane and relative to the anterior direction within the horizontal plane), initial ball velocity, foot velocity, and ball velocity-foot velocity ratio for both types of balls.

RESULTS: The values for initial ball velocity, foot velocity, and ball velocity-foot velocity ratio for all participants are shown in Table1. The mean initial ball velocity and mean ball velocity-foot velocity ratio for the soccer ball were significantly greater than those for the futsal ball. No significant difference was observed between the mean foot velocity when the soccer ball was kicked and that when the futsal ball was kicked. The mean (SD) angles of the pelvis and the thigh-shank plane are shown in Figure1. The mean (SD) changes in the horizontal angles of the pelvis and the thigh-shank plane were similar when both balls were kicked, with no significant difference in the kicking motions.

DISCUSSION: The mean velocity of the soccer ball was similar to those recorded in previous studies on professional soccer players. The results suggested that the kicking performance of professional futsal players is as good as that of professional soccer players. No significant difference was observed between soccer and futsal kicking with regard to the angle of the pelvis and that of the thigh-shank plane. This may indicate that the participants kicked both balls with the same kicking motion. The angles of the pelvis and thigh-shank plane during instep and side-foot soccer kicking have been reported previously. In these studies, the horizontal angle of the pelvis was 105° in instep kicking and 120° in side-foot kicking approximately before the impact, and the horizontal angle of the thigh-shank plane was reported as 92° in instep kicking and 130° in side-foot kicking approximately. The kicking motions of the futsal players were thought to exhibit characteristics of both instep and side-foot kicking, as seen in soccer players. Further, the recorded thigh-shank plane angle also suggested that these subjects kicked the balls at a little inside of the foot. This action results in good impact of the foot on the ball and can be effected by decreasing the positive extension of the foot joint.

Table.1: Initial ball velocities, foot velocities and ball-foot velocity ratios

	Initial ball velocity (m/s)		Foot velocity (m/s)		Ball-foot velocity ratio	
Sub.	Soccer	Futsal	Soccer	Futsal	Soccer	Futsal
A	29.9	27.6	20.5	19.3	1.46	1.43
B	28.3	26.8	19.2	20.7	1.47	1.30
C	30.9	28.2	19.3	20.5	1.60	1.38
D	27.1	24.3	18.6	18.1	1.46	1.34
E	29.7	27.0	20.2	21.7	1.47	1.25
F	28.5	26.1	18.0	18.8	1.59	1.39
G	24.4	23.1	16.2	17.4	1.50	1.32
H	26.3	26.0	18.5	20.0	1.42	1.30
I	27.4	25.4	19.2	18.8	1.43	1.36
Mean	28.1	26.0	18.9	19.5	1.49	1.34
SD	2.0 **	1.6	1.3	1.4	0.06 *	0.06

* Shows significant differnce (*p*<.05)
** Shows significant differnce (*p*<.01)

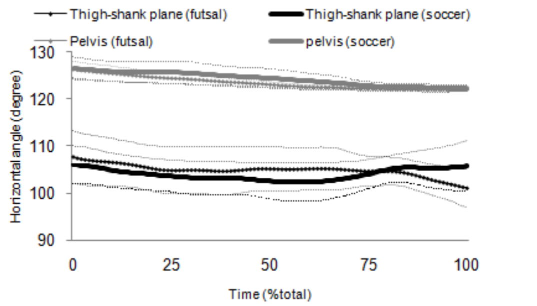

Figure.1: The mean changes and SD for the horizontal angle of pelvis and thigh-shank plane. Time for the leg-aceleration tarm was normalized to 100%.

CONCLUSION: The aim of this study was to clarify the motion involved in kicking a futsal ball. The results show that the kicking performance of professional futsal players is as good as that of professional soccer players. On comparing our results with those of previous studies, we observed that the maximum-effort motion involved in kicking the futsal ball had the characteristics of both instep kicking and side-foot kicking, as seen in soccer players. Futsal coaches training players for powerful kicking should advise the players to use a combination of soccer-style instep kicking and futsal-style side-foot kicking.

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MODEL-BASED IMAGE-MATCHING KINEMATICS ANALYSIS OF THREE ANKLE SUPINATION SPRAIN INJURY CASES IN SPORTS

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Ankle sprain is one of the most common injuries encountered at sport events. Three ankle supination sprain cases from high jump, tennis and hockey were chosen for analysis. Model-Based Image-Matching (MBIM) technique was implemented for reconstructing 3D ankle joint kinematics. The profiles of ankle joint kinematics were outputted from the selected sprain cases. The maximum inversion angle ranged from 78° to 142°. Plantarflexion was again found to be not necessary in ankle supination sprain injury. The results from the MBIM technique would contribute to the understanding of biomechanical injury mechanism of ankle supination sprain injury in sports. The future direction is to analyze more cases to consolidate the findings.

KEYWORDS: ankle sprain, injury biomechanics, video analysis

INTRODUCTION: Ankle sprain is one of the most common injuries encountered at sport events. 20% of the sports injuries are ankle sprains (Fong, 2007). A precise description of the injury situation is a key component to understanding the aetiology and injury mechanism (Bahr, 2005).

Injury incidents are occasionally recorded unintentionally during television broadcasting. Those video recordings provide valuable information for deducing joint kinematics of specific sport injury, as well as, contributing to the study of injury mechanism. From the previous studies, qualitative analysis of joint biomechanics was reported on ankle injuries based on visual inspection (Andersen, 2004). Quantitative analyses on injury cases were available under rare circumstances due to coincidental calibrated video setting (Zernicke, 1977; Fong, 2009). However, body marker tracking system or calibrated video setting are normally not available in all situations. Quantitative description on ankle joint biomechanics was not reported in those studies because no validated method to deduce the three dimensional (3D) joint kinematics from uncalibrated video recordings. In order to develop a novel biomechanical analysis to produce continuous estimates of joint kinematics from video recordings, Krosshaug and Bahr (2005) introduced a Model-Based Image-Matching (MBIM) technique for investigating human motion from uncalibrated video sequences.

The purpose of this paper was to investigate the ankle joint kinematics of three typical ankle sprain cases from take-off stepping, cutting and running motion.

METHOD: Three ankle supination sprain cases from high jump, tennis and hockey were chosen for analysis because of their difference in injury motions and circumstances. The first case was recorded from high jump event in Beijing Olympics Games 2008. The player sprained her left ankle during the take-off stepping. The second case was recorded from male tennis match in Vienna 1995. The player sprained his left ankle during a cutting step with back-hand volley. The third case was captured in male hockey match in Beijing Olympics Games 2008. The player sprained his left ankle during body contact with an opponent. The inclusion criteria were that the player was unable to continue the match or competition after the ankle supination motion, and the injury motion was captured by at least two video cameras. The injury situation and the video information were tabulated in Table 1. The video recordings were transformed from their original format into uncompressed AVI image sequences using Adobe Premiere Pro (version CS4, Adobe Systems Inc., San Jose, California, US). Then the sequences were de-interlaced using Adobe Photoshop (version CS4, Adobe Systems Inc., San Jose, California, US), and the image sequences were synchronized and rendered into 1 Hz video sequences by Adobe AfterEffects (version CS4, Adobe Systems Inc., San Jose, California, US). The matchings were performed using 3D animation software Poser® 4 and Poser® Pro Pack (Curious Labs Inc., Santa Cruz, California, US). The surroundings were built in the virtual environment according to the real dimension of the sport field. The models of surroundings were manually matched to the background for the each frame in every camera view. The skeleton model from Zygote Media Group Inc. (Prove, Utah, US) was used for the skeleton matching. The skeleton was scaled with respect to the height. The skeleton was further scaled with reference to the proportions of body segment which were defined from the standing post image of the player obtained from the internet site. The skeleton matching started with the shank segment and then distally matched the foot, and toe segments frame by frame. The joint angle time histories were read into Matlab (MathWorks, USA) with a customized script for data processing. Joint kinematics was deduced by the Joint Coordinate System (JCS) method (Grood, 1983).The ankle joint kinematics results from MBIM technique were filtered and interpolated by Woltring's Generalized Cross Validation Spline package (Woltring, 1986) with 15Hz cut-off frequency.

The accuracy of MBIM technique on measuring ankle joint kinematics was well-validated (Mok, 2009). The ankle kinematics profile was divided to two phases, pre-sprain and post-sprain, based on the inversion angle.

Table 1. Description of the injury situation and the video sequences

Case no.	Sport event	Gender	Motion	No. of views	Sampling frequency	Video Resolution
1	High Jump (Olympic 08)	Female	Take-off stepping	3	50Hz	1280x720
2	Tennis (Vienna 95)	Male	Cutting	2	50Hz	320x240
3	Hockey (Olympic 08)	Male	Running	2	50Hz	1280x720

RESULT: Figure 1 presents the curve of the ankle joint kinematics of players during the ankle supination sprain injury. Table 2 tabulated the maximum inversion angle, the maximum inversion velocity and the duration from touchdown to maximum sprain. Maximum sprain was defined as the time point of maximum inversion angle reached. The photo captures at the point of maximum inversion angle were shown in Figure 2.

Table 2. The descriptive data of ankle joint kinematics of three injury cases

	Case 1	Case 2	Case 3	Fong et al. (2009)
Max. Inversion angle	142°	94°	78°	48°
Max. Inversion velocity	1752°/s	1488°/s	1397°/s	632°/s
Duration (TD to Max.Sprain)	0.08s	0.12s	0.08s	0.08s

For case one, the player was performing take-off stepping in high jump trial and the injury happened. The maximum inversion angle reached 142° at the point. She twisted her torso for overcoming the bar, leading to her ankle was internal rotated with larger degrees comparing with the other two cases. For case two, the player was performing cutting motion with backhand volley in the tennis match. Sudden plantarflexion was observed at the point of heel touch down. After the full plantar foot contact, increase in inversion and internal rotation were observed until the inversion reach the peak value, 94°. After the peak inversion, the ankle kept internal rotated and plantarflexed. For case three, the player was chasing the opponent with body contact. His left foot slightly stepped on the opponent's foot and the ankle supination motion was triggered. The maximum inversion reached 78°. Internal rotation was observed at the point of peak inversion.

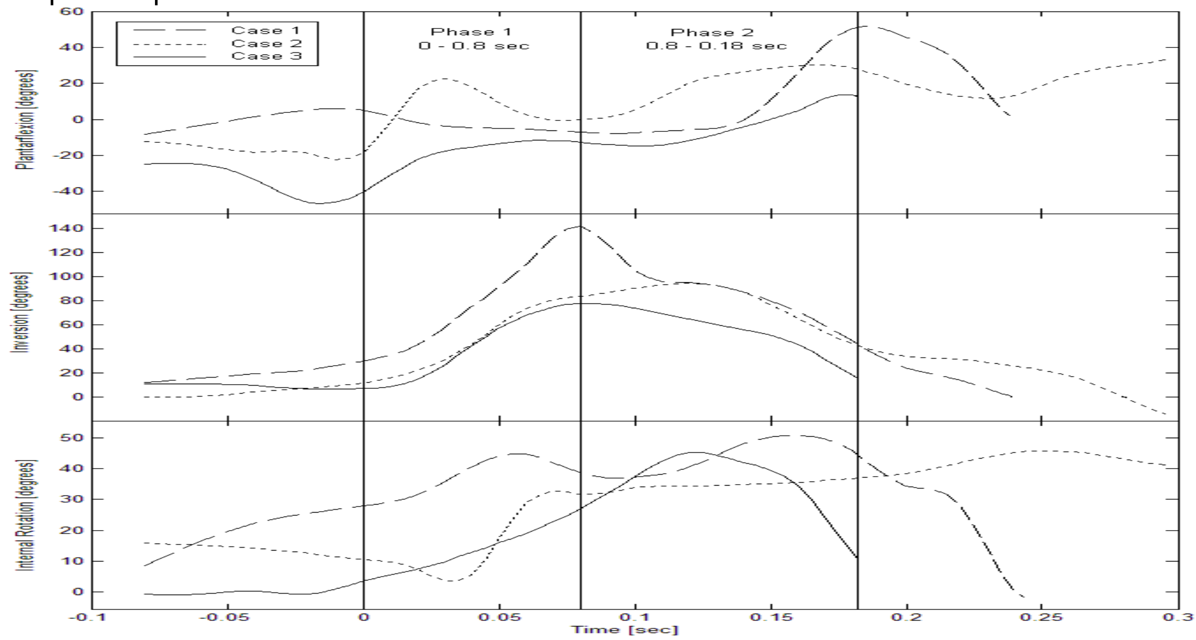

Figure 1. Ankle joint kinematics of the player during the ankle supination sprain injury. Time zero represented the point of heel touchdown. Positive values are Plantarflexion/Inversion/Internal Rotation.

DISCUSSION: Concluding the three selected cases, the ankle kinematics profile were divided into two parts according the phase division approach reported by Fong (2009), shown in Figure 1. Phase 1 was the pre-sprain phase, from 0 sec to 0.8 sec. In this phase, the ankle internal rotation inversion angles were increasing. Phase 2 was the post-sprain phase, the inversion angle decreased and the ankle kept internal rotated.

Compared with the previous biomechanical ankle sprain case report (Fong, 2009), the three analyzed cases were higher in severity. The maximum inversion angle and maximum inversion velocity were higher. The duration from heel touchdown to maximum sprain was similar. Increase in inversion and internal rotation appeared in pre-sprain phase (phase 1).

Large plantarflexion was not found in the analyzed ankle supination sprain injury. There is variation among cases by visual inspection. For instance, vigorous eversion was recorded right after the peak inversion only in case one, however, not in the other two cases.

Figure 2. Captures at the point of maximum inversion angle for Case 1-3 respectively.

CONCLUSION: This study reported the ankle joint kinematics of ankle supination sprain. The results from the MBIM technique could contribute to the understanding of biomechanical injury mechanism of ankle supination sprain injury. Due to variation among cases, the future direction is to analyze more cases to consolidate the findings.

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THE ACUTE TIME COURSE OF CONCURRENT ACTIVATION POTENTIATION

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This study evaluated the acute time course of the ergogenic effect of concurrent activation potentiation (CAP). Forty-two men and women, including CAP non-responders and responders, performed a 5 second isometric knee extension on a dynamometer with the use of remote voluntary contractions (RVC). Mean torque was assessed in seven 500 millisecond (ms) time periods. A two-way repeated measures ANOVA revealed significant main effects for time period ($p \leq 0.001$), but no significant interaction between time period and CAP non-responders and responders ($p > 0.05$). The ergogenic effects of CAP are accrued during the first 1000ms. Concurrent activation potentiation responders produce greater initial force than the CAP non-responders, without a concomitant acceleration in force decay throughout the time course.

KEYWORDS: strength, power, latency, Biodex, motor overflow

INTRODUCTION: Sustaining enhanced muscle performance is the primary goal of strength and conditioning programs. Ergogenic strategies have been sought and the phenomenon of concurrent activation potentiation (CAP) has shown promise. Concurrent activation potentiation is believed to increase prime mover muscle activation during the simultaneous contractions of muscles remote from the prime mover via motor overflow (Ebben, 2006). These simultaneous contractions are defined as remote voluntary contractions (RVC). Previous research assessing the effects of CAP have demonstrated increases in torque during isometric testing (Ebben et al., 2008b) and force development during the countermovement jump (Ebben et al., 2008a). To date, the time course of the ergogenic CAP effect has yet to be examined.

Computerized dynamometry is a commonly used and reliable tool to measure muscular performance of the knee (Surakka et al., 2005) producing variables of power, work, rate of torque development (RTD), and peak torque. Research examining the effects of RVC on peak torque have demonstrated an 8.9% increase in isokinetic knee flexion for men, compared to a condition without the use of RVC (Ebben et al., 2010b). Knee extensors have also been examined isokinetically and have shown peak torque increases of 8.4 (Stumbo et al., 2001) and 10.6% (Ebben et al., 2010b) for men and 4.2% (Ebben et al., 2010b) for women during conditions when RVC were used. Additionally, mean and peak torque increases of 14.6 and 14.8%, respectively, were demonstrated during an isometric knee extension with the simultaneous use of RVC (Ebben et al., 2008b). These studies confirm that torque is greater in the presence of RVC but do not explain the duration of this advantage.

Research evaluating the effect of RVC on RTD and rate of force development (RFD) provide some evidence that the ergogenic effect of CAP may occur early during dynamic movements. Various durations of force development have been used to assess CAP's effect: RTD of the first 300ms (Ebben et al., 2010b), RFD of the first 100ms (Ebben et al., 2010a), RFD of the concentric phase (Ebben et al., 2008a), RFD to peak force (Ebben et al., 2010a), and time to peak force during the concentric phase (Ebben et al., 2008a), all resulting in RVC aided performance increases of 14.6, 32.2, 19.5, 8.2, and 20.2%, respectively. Therefore, it can be hypothesized that CAP's ergogenic effect is accrued quickly.

Thus, the purpose of this study was to assess RVC on mean torque during 500ms time periods of isometric knee extension in CAP non-responders and responders, in order to investigate the time course of the ergogenic effect of CAP.

METHODS: Forty-two subjects participated in this study who were previously classified as CAP non-responders (N=11; mean ± SD, age 20.9 ± 1.9 yr; body mass 69.7 ± 9.8 kg) and responders (N=31; mean ± SD, age 21.3 ± 1.6 yr; body mass 75.7 ± 12.0 kg). Inclusion criteria consisted of at least 2 months of biweekly participation in lower body resistance training with exercises that included knee extension. Subjects were excluded if they presented any history of knee pathology that resulted in functional limitation of the right leg. All participants competed in NCAA Division I athletics, club, or intramural sports, and had played high school sports. Participants refrained from resistance training for at least 48 hours prior to testing. All were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board and designed in accordance with the ethical standards of the Helsinki Declaration.
Subjects warmed up and performed 15 seconds of lower body dynamic stretching for each major muscle group. Subjects were then positioned and secured in a dynamometer (System 4, Biodex Inc., Shirley, NY, USA) according to manufacturer specifications. Straps across the chest, waist, right thigh, and right ankle reduced extraneous movement during the exercise. The right knee joint was positioned goniometrically at 90 degrees and calibrated with the system software and the mass of the limb was measured. Knee joint angle was then adjusted until the software indicated the knee was at 60 degrees of flexion. Test specific warm up sets of two 5 second isometric knee extension exercises at 75 and 90% of their self-perceived maximum ability were then performed. During the exercise, subjects simultaneously used RVC including maximal jaw clenching on a dental vinyl mouth guard (Cramer Products Inc., Gardner, KS, USA), maximal bilateral hand griping using hand dynamometers (model 78010, Lafayette Industries, Lafayette, IN, USA), and the performance of the Valsalva maneuver.
Five minutes of rest were provided between warm up and test sets. Testing consisted of one maximal isometric right knee extension for 5 seconds on the dynamometer with the simultaneous use of the aforementioned RVC. Subjects were instructed to perform maximally.
Isometric torque curves for each subject were analyzed using manufacturer's software. Mean torque was calculated for each of seven 500ms time periods: 0-500, 500-1000, 1000-1500, 1500-2000, 2000-2500, 2500-3000, and 3000-3500, totaling 3.5 seconds. The onset of the 0-500ms time period was identified by the first positive torque value in the torque curve.
All data were analyzed using SPSS 18.0. A two-way repeated measures ANOVA was used to evaluate mean torque, expressed as a percentage of body weight, for the time period and the interaction between the time period for CAP non-responders and responders. Bonferroni adjusted pairwise comparisons were used to assess the specific differences between time periods and CAP non-responders and responders. Statistical power (d) and effect size (η_p^2) are reported and all data are expressed as means ± SD. The *a priori* alpha level was set at $p \leq 0.05$.

RESULTS: The analysis of mean torque, expressed as a percentage of body weight, revealed significant main effects for time course ($p \leq 0.001$, $\eta_p^2 = 0.79$, $d = 1.00$) but no significant interaction between the time course for non-responders and responders ($p = 0.844$). Results of the pairwise comparison are demonstrated in Table 1 and Table 2 in order to provide descriptive data for CAP non-responders and responders, respectively.

Table 1. Mean isometric right knee extension torque as a percentage of body weight (mean ± SD) during 500 millisecond time periods for CAP non-responders (N=11).

0 – 500ms	500 – 1000ms	1000 – 1500ms	1500 – 2000ms	2000 – 2500ms	2500 – 3000ms	3000 – 3500ms
95.38 ± 32.31 [a]	141.16 ± 27.09	147.71 ± 26.21	149.31 ± 25.26	151.49 ± 27.40	153.86 ± 26.71	152.82 ± 25.13

ms = millisecond
[a] Significantly different ($p \leq 0.001$) than all other time periods.

Table 2. Mean isometric right knee extension torque as a percentage of body weight (mean ± SD) during 500 millisecond time periods for CAP responders (N=31).

0 – 500ms	500 – 1000ms	1000 – 1500ms	1500 – 2000ms	2000 – 2500ms	2500 – 3000ms	3000 – 3500ms
105.52 ± 28.77 [a]	154.43 ± 32.14 [a]	163.26 ± 33.27	166.46 ± 33.39	168.41 ± 32.58	169.37 ± 31.38	170.67 ± 30.58

ms = millisecond
[a] Significantly different ($p \leq 0.001$) than all other time periods.

DISCUSSION: This is the first study to investigate the time course of the ergogenic advantage of CAP during isometric knee extension. Non-responders and responders of CAP demonstrated similar torque maintenance during the time course even though the CAP responders mean torque was greater, as shown in previous studies (Ebben et al., 2008a; Ebben et al., 2008b; Ebben et al., 2010a; Ebben et al., 2010b). Practitioners can use this knowledge to prescribe RVC without concern of a higher than usual acute performance decay. This study is the first to demonstrate that the use of RVC do not present a performance decreasing cost to motor resources.

Evaluation of the CAP responder’s mean torque throughout the time course demonstrates the first two 500ms time periods are significantly different than the rest. In the first time period, 0-500ms, the CAP responders demonstrated a 10.1% greater mean torque/ body weight percentage than the CAP non-responders. Collectively, these data potentially explain the performance enhancements in RTD (Ebben et al., 2010b), RFD (Ebben et al., 2008a; Ebben et al., 2010b), and time to peak force (Ebben et al., 2008a; Ebben et al., 2010b). The performance variables assessed in the present, as well as the aforementioned studies, all evaluated force production within the first 500ms or less. The CAP responder’s non-significantly greater mean torque value within the first 500ms of the exercise, compared to the CAP non-responders, is consistent with results from previous research (Ebben et al., 2008a; Ebben et al., 2008b; Ebben et al., 2010a; Ebben et al., 2010b), and together, these data suggest the ergogenic effect of CAP may only occur during the first 1000ms in strength and power tasks.

The present study demonstrates no significant difference in force maintenance between CAP non-responders and responders, with both showing nearly identical percent differences in mean torque values throughout the duration of the time course. Mean torque increased 45.8 and 48.9% from the 0-500ms to the 500-1000ms time period for CAP non-responders and responders, respectively. CAP non-responders and responders again demonstrated similar mean torque increases of 6.6 and 8.8%, respectively, from the 500-1000ms to the 1000-1500ms time period. The comparatively similar linear performance enhancement of the CAP non-responders and responders is consistent throughout the duration of the exercise, suggesting the knee extensor’s performance is not eventually hindered during the use of RVC. Consistent with the present study, previous literature assessing the effect of RVC on isokinetic knee flexor and extensor strength has demonstrated increases in the kinetic knee performance of CAP responders but not CAP non-responders (Ebben et al., 2010b). Therefore concluding, the use of RVC may improve, but do not eventually impede muscular performance.

CONCLUSION: Results of the present study demonstrate CAP's ergogenic effect occurs within the first 1000ms. Concurrent activation potentiation non-responders and responders present no differences in force maintenance, thus the use of RVC do not continually impede muscular performance regardless of the individual's response to CAP. Thus, regardless of an individual's response to CAP, the use of RVC will not adversely affect performance.

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THE EFFECT OF REMOTE VOLUNTARY CONTRACTIONS ON STRENGTH AND POWER TASKS OF WOMEN

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This study evaluated the effect of remote voluntary contractions (RVC's) on the performance of closed kinetic chain exercises. Subjects performed the squat and jump squat in a RVC condition and a condition without RVC's (NO-RVC's). Peak ground reaction force (GRF), rate of force development during the first 100 ms (RFD 100), RFD to peak GRF (RFD-P), and jump squat height (JH) were assessed with a force platform. Data were analyzed with a one way ANOVA. Results revealed there were no significant differences between RVC and NO-RVC conditions for peak GRF for either the squat (p = 0.11) or jump squat (p = 0.47), RFD 100 for either the squat (p = 0.25) or jump squat (p = 0.23), RFD-P for either the squat (p = 0.88) or jump squat (p = 0.38), or for JH for the jump squat (p = 0.68).

KEYWORDS: Concurrent activation potentiation, motor control, ergogenic, training

INTRODUCTION: Concurrent activation potentiation (CAP) has been proposed to enhance prime mover performance via the simultaneous contractions of muscles remote from the prime mover such as jaw clenching (Hiroshi, 2003), which are referred to as remote voluntary contractions (RVC's) (Ebben, 2006). These RVC's have been demonstrated to increase lower body reflexes (Delwaide & Toulouse, 1980; Hortobagyi et al., 2003; Pereon et al., 1995), performance during isometric testing (Ebben et al., 2008a; Sasaki et al., 1998), and the countermovement jump (Ebben et al., 2008b).
A recent review of the literature introduced the concept of CAP and outlined potential methods for optimizing this method of training (Ebben et al., 2006). Since then, researchers studying the effect of RVC's determined that an aggregate of jaw clenching, hand gripping, and the Valsalva maneuver was more effective than jaw clenching or hand gripping alone (Ebben et al., 2008a). The aggregate RVC condition has been shown to produce isometric average torque and peak torque that was 14.6 and 14.8% higher in the RVC compared to the NO-RVC condition (Ebben et al., 2008a). However, given the specious history of the effect of RVC's on muscular performance (Gelb et al., 1996) and the fact that RVC's have been demonstrated to be effective during isometric (Ebben et al., 2008a; Sasaki et al., 1998), but not during some dynamic, tasks (Sasaki et al., 1998). The effect of RVC's on dynamic performance requires further investigation.
Only one published study examined the effect of RVC's during a dynamic athletic event (Ebben et al., 2008b). In this study, subjects produced 19.5% higher RFD and 20.2% faster time to peak force during the countermovement jump while jaw clenching, compared to a non-jaw clenching condition (Ebben et al., 2008b). However, these subjects did not produce greater peak force in the jaw clenching condition. Thus, the potential of RVC's as a potentiation phenomenon for dynamic athletic tasks remains uncertain. The effectiveness of a comprehensive aggregate of RVC's has been demonstrated during isometric testing with men as subjects (Ebben et al., 2008a) but their effect on the performance of dynamic tasks and for women has yet to be investigated. Therefore, the purpose of this study was to

compare conditions that included RVC's and a condition that did not (NO-RVC) and the effect on back squat and jump squat performance as assessed with kinetic data.

METHODS: Subjects included 10 women (mean ± SD, age 20.9 ± 1.1 yr; body mass 65.7 ± 4.41 kg) who participated in intercollegiate or recreational athletics as well as lower body resistance training with exercises that included knee extension for at least 2 months. Exclusion criteria included any history of lower limb pathology that resulted in functional limitation of the exercises to be assessed in this study. The subjects were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board.

Subjects performed a pre-test habituation and test session. Prior to each, subjects warmed up for 5 minutes of light exercise on a rowing ergometer followed by dynamic stretching. A pre-test habituation session was conducted to determine the subjects' 5 repetition maximum (RM) back squat load and countermovement jump height.

During the test session, subjects performed each of the test exercises including 2 repetitions each of the back squat with their previously assessed 5RM load, and the jump squat with an added load equivalent to 30% of the subjects previously estimated 1 RM of their back squat, in the RVC and NO-RVC conditions. In the RVC condition, subjects were instructed to maximally clench their jaw on a dental vinyl mouth guard (Cramer Products Inc., Gardner, KS), grip forcefully on the barbell and pull it down into their trapezius, and perform a brief Valsalva maneuver during the concentric phase of the exercise. The NO-RVC condition included the subjects using their preferred method of gripping the barbell, performing the exercises with an open mouth and pursed lips to limit the likelihood of jaw clenching, and cycling between inspiratory and expiratory flow in order to reduce the Valsalva effect. These methods were similar to those previously used (Ebben et al., 2008a). The order of the test exercises, as well as the order of the RVC and NO-RVC conditions was randomized. Five minutes of rest was provided between each exercise condition to reduce fatigue and order effects. Subjects were instructed to perform maximally and were encouraged equally for all test sets.

The test exercises were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA). The force platform was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Incorporated, Watertown, MA) for later analysis. Peak ground reaction force (GRF), rate of force development during the first 100 ms (RFD 100), RFD to peak GRF (RFD-P), and jump squat height (JH) were calculated from the force-time records consistent with methods previously used (Jensen & Ebben, 2007). All values were determined as the average of 2 trials for each exercise. Peak GRF during the concentric phase was defined as the highest value attained. The RFD-100 and RFD-P were defined as the first peak of GRF minus the initial GRF during the concentric phase divided by the time to the first peak of GRF minus the time of initial GRF, and normalized to one second. These two RFD measures were calculated consistent with the methods used by Jensen et al. (2008) and were used to assess faster and slower components of the stretch shortening cycle based on the concept proposed by Schmidtbleicher (1992).

All data were analyzed with SPSS 16.0 using a one way ANOVA to evaluate the differences between the RVC and NO-RVC conditions. Statistical power (*d*) and effect size (η_p^2) are reported and all data are expressed as means ± SD. The *a priori* alpha level was set at $p \leq 0.05$.

RESULTS: There were no significant differences between RVC and NO-RVC conditions for peak GRF for either the squat ($p = 0.11$) or jump squat ($p = 0.47$). There were no significant differences between RVC and NO-RVC conditions for RFD 100 for either the squat ($p = 0.25$) or jump squat ($p = 0.23$). There were no significant differences between RVC and NO-RVC conditions for RFD-P for either the squat ($p = 0.88$) or jump squat ($p = 0.38$). There were no significant differences between RVC and NO-RVC conditions for JH for the jump squat (*p* = 0.68). Data for the squat and jump squat are presented in Table 1.

Table 1. Data presented as mean ± SD for the squat and jump squat for subjects in the RVC and NO RVC conditions.

	Squat			Jump Squat		
	RVC	NO-RVC	%	RVC	NO-RVC	%
Peak GRF (N)	1546.2±186.2	1501.8±181.4	2.9	1422.5±106.7	1417.7±106.8	0.03
RFD 100 (ms)	1004.1±680.8	1517.7±1050.9	-33.9	1707.4±1286.7	2101.9±1749.2	-18.8
RFD to peak	255.2±127.7	261.8±125.2	-2.5	797.2±256.0	840.7±144.2	-5.2
Jump Height (m)	--	--	--	0.13±0.04	0.13±0.03	0.0

DISCUSSION: This is the first study to investigate the effects of CAP during ground based exercises such as the squat and jump squat with women subjects, demonstrating that subjects in the RVC condition, compared to the NO-RVC condition, accrued no statistically significant higher performances for any of the outcome variables assessed.
Previous research examining the effects of RVCs during the squat and jump squat revealed that men accrued a statistically significant advantage in the RVC compared to the NO-RVC condition (Ebben et al., in press). Given the similarity between the studies but differences in the results, the present study raises questions about the effectiveness of RVC's for women subjects.
Previous research using only men as subjects (Ebben et al., 2008a) examined the effect of CAP during ground based closed kinetic chain exercise and demonstrated 19.5% higher RFD during the countermovement jump while in the RVC compared to the NO-RVC condition and higher RFD for a subject sample that included both men and women, though no separate gender bases analysis was performed (Ebben et al., 2008b).
In the present study, the mean RFD values were appreciably lower in the NO-RVC condition, though large standard deviations suggest significant variability in subject ability.

CONCLUSION: Results of this study demonstrate no statistically significant performance increase in the variables assessed when using RVC's. In some cases, the use of RVC's resulted in a non statistically significant decrease in performance. In contrast to other studies, RVC's may only augment performance in men, but not women.

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THE EFFECT OF ANTAGONIST CONDITIONING CONTRACTIONS ON LOWER AND UPPER BODY POWER TESTS

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This study assessed the effect of antagonist conditioning contractions (ACC) on lower and upper body power tests. Six subjects performed the bilateral countermovement jump and the supine medicine ball chest throw on a force platform in baseline conditions and after ACC. A repeated measures ANOVA was used to compare performance between the baseline conditions and the ACC condition. Analysis demonstrated no significant main effects for GRF (p = 0.41) or RFD (p = 0.55) for the countermovement jump. Additionally, there were no significant main effects for GRF (p = 0.85) or RFD (p = 0.95) for the medicine ball throw. This study demonstrated that maximal short term ACC do not enhance multi-joint power tests such as the countermovement jump and medicine ball throw.

KEYWORDS: reversal of antagonists, successive induction, Golgi tendon organ, superset

INTRODUCTION: The activation of the antagonist in order to potentiate the agonist muscle group has been thought to stimulate the Golgi tendon organ (GTO) (Kroll, 1972). Golgi tendon organ activation may inhibit the antagonist and activate the agonist. This process has been termed successive induction, the reversal of antagonists, and is typified by the rapid transitions between antagonist and agonist muscle groups during movements such as walking, running, and rowing (Voss et al., 1985). Additionally, skilled athletes appear to be able to reduce antagonist co-activation as an adaptation that allows them to produce greater agonist force (Bazzucchi et al., 2008).
A small body of literature also defines phenomena such as successive induction and the reversal of antagonists, as antagonist conditioning contractions (ACC). Research examining the role of ACC on agonist force demonstrates higher rates of force development (RFD) but neither higher peak force, nor increased muscle activation (Gabriel et al., 2001; Grabiner et al., 1994; Kamimura et al., 2009) during single joint strength tasks. Furthermore, some evidence demonstrates that stimulating a muscle with a maximal or near maximal activation may potentiate rather than inhibit it (Hodgson et al., 2005) and efforts to reduce the activation of the antagonist, via a fatiguing stimulus, resulted in its potentiation and subsequent impairment of agonist functioning (Maynard and Ebben, 2003). Thus, the challenge seems to be to activate the antagonist enough to stimulate the GTO, while not potentiating it, and to take advantage of the antagonist inhibition before it decays.
Furthermore, if this process has ergogenic value and external validity it would need to translate to functional movements. Therefore, the purpose of this study was to assess the effect of ACC on lower body power during the countermovement jump and upper body power during supine medicine ball chest throws.

METHODS: Six men (mean ± SD: age = 21.50 ± 1.05 years; height = 174.78 ± 7.56 cm; body mass = 79.54 ± 13.72 kg; countermovement jump height = 67.75 ± 7.24 cm; seated 6 kg medicine ball throw = 427.83 ± 43.85cm) volunteered to serve as subjects for the study. Inclusion criteria required subjects who were 18-27 years old and participated in upper and lower body resistance training for at least 8 weeks prior to testing. Exclusion criteria included any orthopedic upper or lower limb or cardiovascular pathology preventing maximal effort. Subjects signed an informed consent form prior to participating in the study which was approved by the Institutional Review Board.

Subjects warmed up and then participated in a single 60 minute testing session. The warm up consisting of 3 minutes on a rowing ergometer followed by a 10 yard forward walking lunge, 10 yard backward walking lunge, 10 yard walking hamstring stretch, 10 yard walking quadriceps stretch, 5 slow bodyweight squats, 5 fast bodyweight squats, 5 small arm circles forward, 5 small arm circles backward, 5 large arm circles forward, and 5 large arm circles backward. Five medicine ball chest throws and 5 countermovement jumps were performed at 75% of the subject's self-assessed maximal effort. Then, two maximal repetitions of each were performed and the average was recorded. These medicine ball chest throws and countermovement jumps were performed in order to evaluate and demonstrate subject characteristics. The vertical jump was assessed with a Vertec. The medicine ball chest throw was performed from a seated position with the subject's back against a wall using a 6 kg medicine ball. Subjects then rested for 4 minutes.

During the test, subjects performed 6 sets including 3 sets of 3 repetitions of countermovement jumps and supine medicine ball chest throws. The 3 test sets of each exercise were performed in the following order, including a pre ACC baseline condition with no preceding ACC, an ACC condition with a preceding ACC, and a post ACC baseline condition without a preceding ACC. The order of the test exercises (countermovement jump or medicine ball chest throw) was randomized and kept consistent throughout all 3 test sets. Four minutes of rest was used between the pre ACC baseline condition and the ACC condition. Fifteen minutes of rest was provided between ACC and post ACC baseline test in order to reduce the potential potentiation of the antagonist from the previous set since the potentiation effect is believed to be brief (Chalmers, 2004). Thus, testing included 2 non ACC baseline conditions with one before and one after the ACC condition in order to minimize potential order effects. Merely randomizing the ACC and non ACC conditions might have resulted in residual ACC potentiation in the non ACC conditions when randomization resulted in the non ACC conditioning occurring after the ACC condition.

The countermovement jump was performed from a standing position on a force platform whereas the supine medicine ball chest throw was performed on a flat utility bench on the force platform. During the supine medicine ball chest throw subjects were positioned with shoulders flexed to 90°, elbows extended, and their head, upper and lower back, buttocks, and both feet on the utility bench. A 9 kg medicine ball was placed in the hands of the subject. Following a verbal cue, the subject performed a countermovement and then explosively threw the medicine ball vertically. Researchers caught the ball and re-administered the medicine ball to the subject for the next repetition. The pre and post ACC baseline conditions were performed without an ACC. The ACC condition consisted of a brief 6 second isometric contraction for the hamstrings and the shoulder retractors for the bilateral countermovement jump and supine medicine ball chest throw, respectively. Following the cessation of the 6 second ACC of each exercise, subjects were allotted 10 seconds to reposition themselves for the respective test set.

The test exercises were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA, USA) which was bolted to the laboratory floor and mounted flush in the center of a 122 x 244 cm weightlifting platform. The force platform was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA USA) for later analysis.

Vertical ground reaction force data were acquired and reduced with manufacturer software Peak vertical ground reaction force (GRF) and RFD was calculated from the force-time records for the concentric phase. Peak GRF was defined as the highest value attained from the force time record. The RFD was defined as the peak GRF minus the GRF value 100 ms prior to the peak divided by 100 ms, consistent with methods previously used (Ebben et al., 2008). Three repetition averages were then calculated. Data were evaluated with SPSS

18.0 for Windows (Microsoft Corporation, Redmond, WA, USA) using a repeated measures ANOVA to determine statistical differences in GRF and RFD between the test conditions. Assumptions for linearity of statistics were tested and met. All data are expressed as means ± SD.

RESULTS: Analysis demonstrated no significant main effects for GRF (p = 0.41) or RFD (p = 0.55) for the countermovement jump. Additionally, there were no significant main effects for GRF (p = 0.85) or RFD (p = 0.95) for the supine medicine ball chest throw. Mean data are presented in Table 1-4.

Table 1. Mean (± SD) concentric peak vertical ground reaction force (GRF) and rate of force development (RFD), during the countermovement jump (CMJ) in the pre and post antagonist conditioning contraction (ACC) baseline conditions, and during the ACC condition.

	PRE	ACC	POST
CMJ GRF (N)	1246.95 ± 429.79	1406.88 ± 295.47	1368.17 ± 270.43
CMJ RFD (N·sec^{-1})	4335.36 ± 1729.90	4800.94 ± 1356.92	4301.57 ± 1467.15

Table 2. Mean (± SD) concentric peak vertical ground reaction force (GRF) and rate of force development (RFD) during the medicine ball (MB) chest throws in the pre and post antagonist conditioning contraction (ACC) baseline conditions, and during the ACC condition.

	PRE	ACC	POST
MB GRF (N)	661.58 ± 107.47	686.88 ± 163.06	692.82 ± 208.98
MB RFD (N·sec^{-1})	6003.61 ± 1223.85	6166.00 ± 1578.38	5912.94 ± 2586.80

DISCUSSION: This study shows that ACC do not significantly increase performance in lower or upper body power exercises, despite using brief and maximal ACC based on previous recommendations (Grabiner et al., 1994; Holt et al., 1969; Kamimura et al., 2009). These results raise questions about the effectiveness of activating the antagonist in order to augment performance of a subsequently activated agonist during multi-joint movements.

Results of the present study differ from previous research examining the effect of ACC which demonstrated increased RFD (Gabriel et al., 2001; Grabiner et al., 1994; Kamimura et al., 2009), but not force (Grabiner et al., 1994; Kamimura et al., 2009) or work (Grabiner et al., 1994) during single joint and less dynamic tasks.

Most studies assessing ACC failed to find any increase in electromyography of the agonist (Gabriel et al., 2001; Holt et al., 1969; Kamimura et al., 2009) demonstrating that either EMG was unable to detect, or another mechanism was responsible for, the increase rate of force development demonstrated in these studies. This absence of statistically significant differences in the present study may be due to the fact that the ACC mediated inhibition may only last one second (Chalmers, 2004). Thus, it is possible that the transition from the ACC to medicine ball chest throw or countermovement jump may not have occurred within the 1 second time interval. While ACC may not work well for strength and power tasks, a functional benefit may still be present during the reversal of antagonists for a variety of functional movements that quickly transition between antagonist and agonist such as walking or running (Voss et al., 1985) or as a potential for chronic adaptation in skilled performers (Bazzucchi et al., 2008).

In the present study, for the countermovement jump, the ACC was performed for the hamstring muscle group. While the hamstring muscle group is a knee extensor antagonist, it is a hip extensor agonist which is used during the countermovement jump.

It should be noted that, despite no statistically significant differences between the baseline and the ACC conditions, the mean countermovement jump RFD and GRF is at least 10.7% and 2.77% higher, respectively, during the ACC compared to the baseline condition. For the medicine ball chest throw, the mean RFD is 2.72% higher. These mean data are consistent with previous reports that ACC may have a larger effect on RFD than peak force production (Gabriel et al., 2001; Grabiner et al., 1994; Kamimura et al., 2009). In the present study, large subject variability, manifested as large standard deviations, and a small number of subjects may have precluded a finding of significance. Thus, the use of ACC during dynamic multi-joint tasks should be further investigated.

CONCLUSION: This study demonstrated that maximal short term ACC do not enhance multi-joint power tests such as the countermovement jump and medicine ball throw.

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DETERMINATION OF BODY SEGMENT INERTIA PARAMETERS USING 3D HUMAN BODY SCANNER AND 3D CAD SOFTWARE

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KEYWORDS: BSP, Solid model, Athlete.

INTRODUCTION: In the field of sports biomechanics, a human body is often treated as a linkage model to investigate various kinds of human movement. This modeling requires body segment inertia parameters (BSPs) such as masses, centers of mass, and moments of inertia. As the quality of motion capture system increases, more accurate BSPs are also needed to get accurate inverse dynamics results. Advanced technology has enabled us to obtain three-dimensional coordinates of the entire body surface. A 3D CAD software has also been able to be applied to measure the human body. It was hypothesized that BSPs with high accuracy could be determined by the combination of a 3D body scanner and a 3D CAD software. The purposes of this study are, first, to introduce a new method of measuring subject-specific BSPs and, second, to compare the BSPs from this study with those from an existing mathematical model in order to show that the proposed method can be used to produce more accurate BSPs.

METHOD: Subjects were 14 Japanese elite male distance runners (22.1±0.9 yr.). An optical body scanner (Hamamatsu Photonics K.K., Japan) was used to acquire three-dimensional surface coordinates of a standing body at 2.5mm height intervals. It took only 13 seconds to measure the coordinates with about 150,000 points over the entire body surface. The measured coordinates were then imported into a 3D CAD software (Dassault Systèmes SolidWorks Corp., USA) to produce a solid model that treats each segment as a rigid body. The procedure to produce the solid model and to calculate its BSPs is as follows.

1. A mesh was made from the point coordinates (Fig. 1 ①→②).
2. The mesh was smoothed (Fig. 1 ②→③).
3. A solid model was configured from the mesh (Fig. 1 ③→④).
4. The solid model of one body segment was cut out (Fig. 1 ④→⑦).
5. Its volume and the preliminary moments of inertia about its three principal axes were calculated by a modeling kernel.

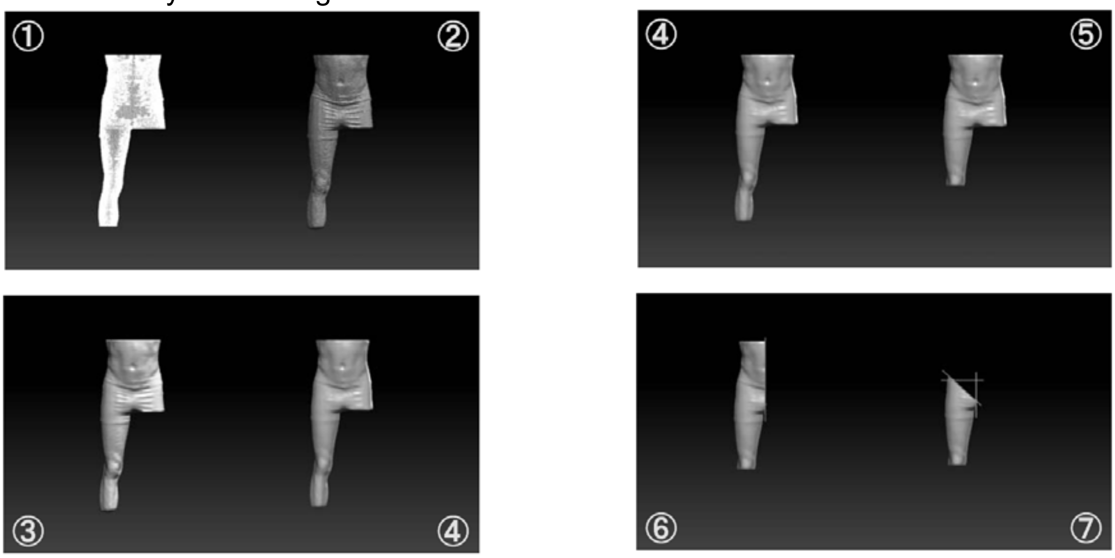

Fig. 1 Procedure for producing our solid model.

This procedure was repeated for each of the 14 segments (head, torso, upper arms, forearms, hands, thighs, shanks, and feet). Each segment density was assumed to be uniform. The optimal density set of the 14 segments for each subject was selected from the 26 density sets of cadavers (Dempster, 1955; Chandler et al., 1975). Subsequently, the density of each segment was adjusted in such a way that the sum of products of the calculated volume and the optimal density for each segment corresponds to the whole body mass. Finally, the mass, the center of mass, and the moments of inertia about the three principal axes for each segment were determined from the calculated volume, the preliminary moments of inertia, and the adjusted density.

RESULTS & DISCUSSION: We were able to construct a high shape fidelity body model by using the 3D body scanner and the 3D CAD software. Figure 2 illustrates our solid model and an exsisting elliptical-cylinder model. It shows that the solid model can reconstruct the body contour and shape faithfully. In contrast, the elliptical-cylinder model is too simplified to get an unevenness in details of a body part. To compare the two models, volumes of all 14 body segments were examined. When the elliptical-cylinder model was used, torso volume were 5% overvalued and the other body segments volume were 5-30% undervalued. Because the body shape is faithfully reproduced, our solid model has advantages over currently used elliptical-cylinder models for measuring subject-specific BSPs accurately. In addition, our new method is less demanding on subjects, less time-consuming for them like the photogrammetry.

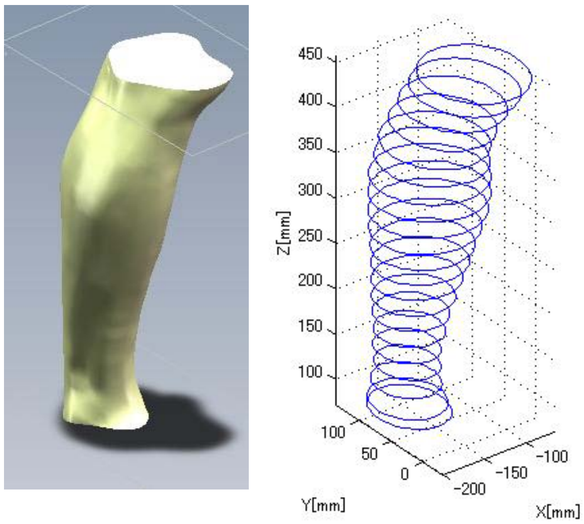

Fig. 2 Our solid model and a currently used elliptical-cylinder model.

CONCLUSION: This study proposed a new method of measuring subject-specific BSPs by using the 3D human body scanner and the 3D CAD. The measured values by our method have been thought more accurate than those measured by an existing elliptical-cylinder model.

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THE EFFECT OF REACHING TO AN OVERHEAD GOAL WHILE PERFORMING THE COUNTERMOVEMENT JUMP

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One potentially simple way to maximize jumping effort and thus intensity is to have athletes jump to and attempt to touch challenging overhead goals during training. The purpose of this study was to compare the effect of jumping with and without the use of an overhead goal. Subjects performed 3 countermovement jumps in conditions with and without an overhead goal. Jump performance was evaluated using a force platform to determine peak ground reaction force, time to takeoff, power, and jump height. Data were evaluated with a two way ANOVA with results demonstrating no significant (p > 0.05) difference between goal conditions for any of the variables assessed and no interaction between goal condition and gender (p > 0.05).

KEYWORDS: plyometrics, effort, intensity, external focus

INTRODUCTION: Research has evaluated the intensity of plyometric exercises indicating that exercise intensity is an important program design variable (Ebben et al., 2008; Jensen & Ebben, 2007). In an attempt to optimize plyometric training intensity, studies have examined internal and external focusing strategies including the use of overhead goals in an effort to examine their effect on jumping performance (Edwards et al., 2008; Ford et al., 2005; Tod et al., 2009; Wulf et al., 2007; Wulf & Dufek, 2009).

Parameters of vertical jump performance including center of mass displacement, impulse, and joint angular velocity have been improved through the use of self talk or exogenous instruction, compared to control conditions (Edwards et al., 2008; Tod et al., 2009), without any reported change in jump height.

Studies evaluating the use of an overhead goal such as jumping to a suspended ball demonstrate slightly higher jumps of approximately 2.2% when compared to a condition without the overhead goal. However, this difference was present only for men without any significant differences for women (Ford et al., 2005). The men in this study accrued this advantage without any difference in jumping biomechanics between the conditions (Ford et al., 2005). When a Vertec (Vertec 2, Sports Imports, Columbus OH) jump height measurement device was used as an overhead focus point, subjects produced higher center of mass displacement, impulse, and joint angular velocities compared to test conditions of using the fingers as focusing strategy (Wulf, et al.,2007; Wulf & Dufek, 2009). Numerical data are limited and the effect of using the Vertec as an external focusing strategy is only compared to the internal strategy of focusing on the fingers while jumping to touch the vanes of the Vertec in some cases (Wulf & Dufek, 2009). Thus, it is unclear how jumping to an overhead goal of the Vertec would differ from a condition of no overhead goal at all. In applied strength and conditioning settings, frequently athletes perform plyometric exercises without any use of an overhead goal. The purpose of this study was to compare the effect of performing the countermovement jump with and without the use of an overhead goal by evaluating subject peak vertical takeoff ground reaction force (GRF), time to takeoff (TTT), power (P), and jump height (JH).

METHODS: Twenty one current or former NCAA Division I and Division II athletes (15 men mean ± SD; age = 22.9 ± 5.9 years, height = 180.0 ± 8.0 cm; body mass = 85.58 ± 9.06 kg) and 6 women; mean ± SD; age = 21.2 ± 1.5 years, height = 168.05 ± 7.9 cm; body mass = 66.27 ± 6.8 kg) served as subjects. All subjects performed a habituation and testing session. Prior to testing, the subjects warmed-up and performed dynamic stretching and jumping exercises. During habituation, subjects were given instruction, a demonstration, and practiced the correct performance of the countermovement jump with and without the Vertec as an overhead goal. During this time, the countermovement jump height was assessed so this value could be used during the test for the overhead goal condition. During testing, subjects performed 6 repetitions of the bilateral countermovement jump with 3 repetitions each alternating repetition by repetition between the jumping to overhead goal and the non-overhead goal conditions. The initial starting order was counterbalanced between subjects. Subjects were allowed to rest 1 minute between test repetitions.

The test exercises were assessed with a force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA, USA) which was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Incorporated, Watertown, MA USA) for later analysis. All values were averaged for three trials for each plyometric exercise.

Peak vertical GRF during takeoff, TTT, P, and JH were calculated from the force time records consistent with methods previously used (Canavan & Vescovi 2004; Jensen & Ebben 2007; Moir 2008; Raynor & Seng 1997; Tsarouches et al., 1995, Van Soest et al., 1985). Peak GRF was defined as the highest value attained from the force time record for the take off phase of each jump. Time to takeoff was defined as the time of the point of onset of the flight phase minus the point of onset of the eccentric phase from the force time record. The point of onset of the eccentric phase was identified consistent based on methods previously demonstrated by Jensen et al. (2009). Jump height and power were calculated based in part on flight time using previously published equations (Moir, 2008).

The statistical analyses were undertaken with SPSS 17.0 using the average of the three repetitions of the test countermovement jumps. A two way mixed ANOVA was used to evaluate the main effects for overhead goal condition and the interaction between overhead goal condition and gender for GRF, TTT, P, and JH. The trial to trial reliability of each dependent variable was assessed for each overhead goal condition using average measures intraclass correlation coefficient (ICC). In addition, a repeated measures ANOVA was used to confirm that there was no significant difference ($P > 0.05$) between three trials of each overhead goal condition. An *a priori* alpha level of $P \leq 0.05$ was used.

RESULTS: There was no significant main effects for jump condition for GRF ($p = 0.22$), TTT ($p = 0.17$), P ($p = 0.70$) or JH ($p = 0.70$). There was no significant interaction of jump condition and gender for GRF (p= 0.39), TTT ($p = 0.98$), P ($p = 0.80$) or JH ($p = 0.79$). Trial to trial reliability of all measures in both the overhead goal and no-overhead goal conditions were highly reliable as demonstrated by ICC values of .96 to .99 with no significant differences between trials ($p > 0.05$). Data are presented in Table 1.

Table 1. Mean ± SD ground reaction forces (GRF), time to takeoff (TTT), power (P), and jump height (JH) in overhead and non overhead goal conditions for men and women

	Men (N=15)		Women (N=6)	
	Goal	No Goal	Goal	No Goal
GRF (N)	1383.14 ± 256.12	1349.15 ± 259.89	807.88 ± 95.28	802.58 ± 106.19
TTT (s)	0.62 ± 0.19	0.64 ± 0.16	0.61 ± 0.07	0.62 ± 0.16
P (w)	4379.69 ± 557.19	4382.95 ± 505.27	2862.14 ± 475.12	2879.82 ± 456.13
JH (m)	0.44 ± 0.07	0.44 ± 0.06	0.32 ± 0.06	0.32 ± 0.06

DISCUSSION: This study demonstrates no significant difference between jumping to an overhead goal versus not jumping to an overhead goal for both men or women. This finding

was not consistent with the previous findings of small performance differences between overhead goal and no overhead goal conditions (Ford et al., 2005; Wulf et al., 2007; Wulf & Dufek, 2009) and results are similar to the absence of any gender differences in jumping to overhead goals as previously demonstrated (Ford et al., 2005).
Studies have also conceptualized this process of training with overhead goals based in the nature of the focusing strategy including an external focus on the overhead goal and an internal focus on the fingers that touch the goal. In some cases, higher performances when using an external focus compared to the internal focus condition was not compared to a non-overhead goal condition so it makes interpretation of the benefits of an overhead goal difficult to assess (Wulf & Dufek, 2009).
Subject training status and experience with plyometric exercises may affect the potential value of an external overhead goal. In the present study, experienced and well trained subjects may have been more able to give high quality effort without an external goal whereas those who are less experienced or motivated may perform better with the use of the goal. Unfortunately, previous studies do not provide measures of their subjects training status. One study reported using "Division I" soccer players who jumped and grabbed a suspended ball placed overhead. However, presumably soccer players are not experienced with this type of athletic movement since grabbing the ball is against the rules of the sport. Therefore, this is an atypical movement for these athletes and may add a confounding variable to this test (Ford et al., 2005).

CONCLUSION: Performing plyometric exercises with maximal effort it likely to be important along with the progression of exercise intensity. However, based on the kinetic variables assessed in this study the use of overhead goals neither improves nor impairs maximum jumping in fairly well trained subjects.

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A New Approach for Assessing Kinematics of Torso Twist in Baseball Batting: A Preliminary Report

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KEYWORDS: projection angle, proximal-distal pattern, pelvis, thorax, shoulder girdles

INTRODUCTION: The motions of segments involved in striking and throwing events are generally sequenced in a proximal-to-distal fashion (Putnam 1993). Welch et al. (1995) analyzed baseball batting using a rigid-body-link model with a two-segment torso, and indicated the importance that the lower torso starts rotation in the direction of pitcher before the upper torso, which, in turn, should start before the arm segments. Also, this sequential motion is considered to allow the kinetic link system to generate synergy between the musculature of the torso and upper extremity. Specifically, the upper torso is expected to have an important function to accelerate distal upper extremity and bat. However previous studies employed a two-segment torso model and the influence of the motions of the shoulder-girdles to the torso's sequential action for twisting was ignored. In the present study, torso's sequential twisting was analyzed with a three-segment torso model, and the kinematics of torso twist in baseball batting was evaluated.

METHOD: Two male collegiate baseball batters (height: 175.3 ± 2.5 cm, mass: 65.1 ± 0.2 kg, age: 19 yr) participated in this study. After performing sufficient warm-up, they engaged in a series of batting trials using a batting tee. Each subject was asked to adjust the horizontal position of the batting tee and the height of the tee was set at the level of his anterior superior iliac spines (ASISs). Each subject was instructed to hit the ball towards center field with maximal effort. The experiment was continued until three successful batting trials were recorded with an optical motion capture system (VICON-MX, Oxford Metrics Inc.) at the sampling rate of 500Hz. The positions of acromions (ACs), ASISs, sternal notch (SN), xiphoid process (XP), seventh cervical spine (C7), eighth thoracic spine (T8), ASISs, and posterior superior iliac spines (PSISs) were measured in the global coordinate system (Gs) to formulate a three-segment torso model. For the trial in which the fastest bat-head velocity was attained, the orientations of the upper, middle, and lower torso segments were determined. The orientations of the lower torso (pelvis) and the upper torso (shoulder girdles) were defined as the horizontal projections of the vector connecting ASISs and of the vector connecting ACs, respectively (Fig. 1). The orientation of the mid-torso (throax) was defined as the horizontal projection of the cross-product computed from the vector pointing from C7 to SN and the vector pointing from center of SN and C7 to the center of XP and T8. The orientation of each segement was then expressed as the angle measured from the Y-axis. The angular velocity of each segment was also calculated as the time derivative of the angle.

The accuracy in defining the three-segment torso model with the present approach was assessed by determining the root mean square errors (RMSE) of the lengths and the angles of the quadrangles representing the segments of thorax and pelvis computed over the analysis interval (200 ms).

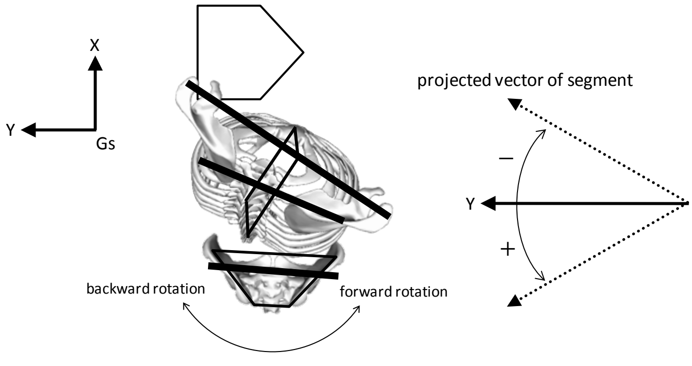
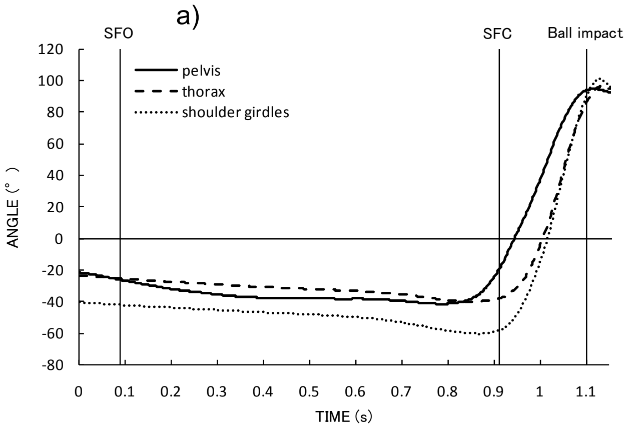
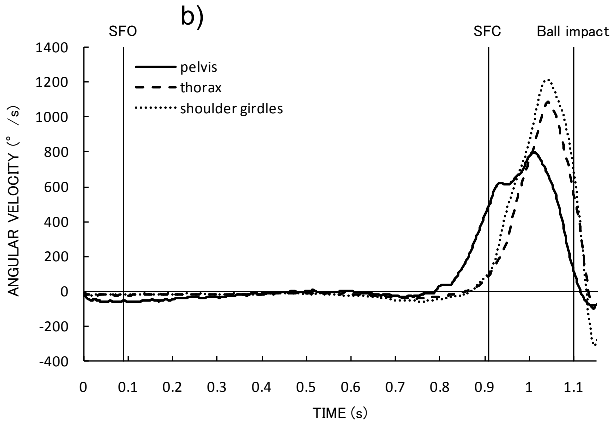


Figure 1. Over-head of the three-segment torso model (left figure) and definition of the rotation angle (right figure) for a right-handed

RESULTS: The accuracy of the three-segment torso model was reasonable for the present study (RMSE of 0.3 cm and 0.5° for pelvis segement and 0.4 cm and 1.5° for thorax segment, respectively). At step foot off (SFO), the orientations of the pelvis and thorax were angled to face slightly backward by a similar amount (-25 ± 1° and -21 ± 1°, respectively) and that of the shoulder girdles was angled even more in the same direction (-35 ± 7°). Shortly before the instant of step foot contact (SFC), the pelvis initiated rotation, and the thorax and shoulder girdles followed it simultaneously at SFC (Fig. 2a). The ranges of motion were approximately 120° for the pelvis and thorax and 144 ± 7° for the shoulder girdles. The pelvis recorded the peak angular velocity at 1.012 s after SFC (Fig. 2b), and 0.028 s later the thorax and shoulder girdles attained the peaks (1165 ± 49°/s and 1007 ± 78°/s, respectively).





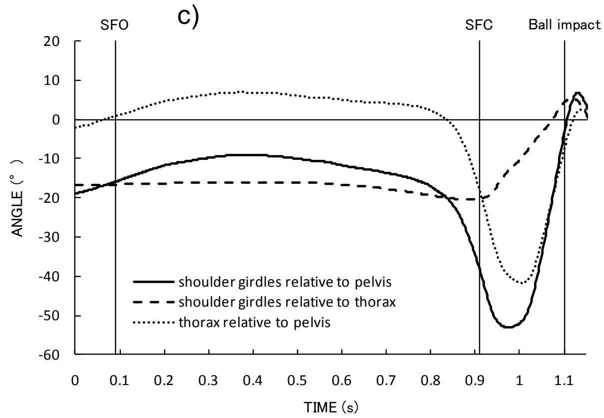


Figure 2. A typical pattern of change in angular displacement (a) and velocity (b) of pelvis, thorax and shoulder girdles, and the relative angles (c) of each segment.

DISCUSSION: The difference in the angles between the pelvis and thorax indicates the twisting of the thorax relative to the pelvis (Fig. 2c). The accumulated twist up to 0.1 s prior to the ball impact was returned completely by the ball impact. This accumulation-return of twist might produce a stretch-shortening effect. The delayed timings at which these segments attained the maximum angular velocity indicate that the pelvis and thorax move with the proximal-distal pattern. In contrast, no difference in timing was found between the thorax and shoulder girdles suggests that the proximal-distal pattern, and also the stretch-shortening effect, does not exist between these segments.

CONCLUSION: With the new kinematic model, the pelvis and thorax were found to move in a proximal-distal sequence, and their twist-accumulation was followed by immediate return observed within 0.2 s suggests an involvement of stretch-shortening cycle. The thorax and shoulder girdles did not demonstrate such patterns during baseball batting.

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KINETICS OF DODGEBALL THROWING WITH AN IMPLICATION ABOUT INJURY MECHANISMS OF ELBOW JOINT

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The purpose of this study was to investigate the load of elbow joint based on the kinetic analysis of the primary school dodgeball players' throwing motion. Four male primary school dodgeball players performed the overhand throwing motion while recorded by a Vicon three-dimensional motion analysis system. Changing patterns of elbow joint torques are similar to those for baseball pitching. The varus torque was the largest value among all elbow joint torques . The extension/flexion angle of elbow joint was -64.8° at peak varus torque. It is possible that the varus torque may lead to medial elbow injury, including ligament, muscle, joint surface and ulnar nerve damage. For healthy growth of children, it would be necessary to set some rules for preventing acute and chronic sports injuries, for example, the number of times of throwing in day should be limited.

KEYWORDS: dodgeball, kinetics, injury, primary school, throwing.

INTRODUCTION: Dodgeball is one of the most popular recreational and competitive activities for primary school children in Japan (Wang et al., 2008). The mass of the ball for primary school children is 380g, and is nearly three times of the mass of a baseball. The speed of the ball being thrown is considered to be the most important factor to win the game. It can be estimated that the load at arm joint would be the larger with the larger mass or the faster speed of the ball. The basic method of throwing is overhand, which is similar to the overhand pitching of baseball. The injury in elbow joint is the most frequently occurred injury among the baseball players by the reason of varus torque applied in the elbow joint (Atwater, 1979; Fleisig et al., 1995). We can guess that it is possible for the primary school dodgeball players to suffer the same injury in elbow joint just like baseball players. The purpose of this study was to investigate the load of elbow joint using a kinetic analysis of primary school dodgeball players' throwing motion.

METHOD: Four male primary school dodgeball players (Height=1.37±0.05m, Body mass=30.1±3.2kg, Age=10±0 years, 2-3 years of experience) were used as subjects for this study. All subjects were right-handed with overhand throwing style.
Based on the provision about the ethical guidelines for research on human established by the Ethics Committee of Graduate School of Health and Sport Sciences of Chukyo University, the purpose and the details of this experiment had been explained to the subjects and their protectors before the experiment. The permission to participate the experiment were collected from subjects and the protectors after understanding the details of this study. None of the subjects was found being injured by medical examiner before the throwing experiment. After ordinary warm-up, reflective markers were attached to the skin of anatomical landmarks of the subjects' body, namely, the head of the 3rd metacarpal on the dorsal of the throwing

hand, most caudal-medial point on the ulnar/radial styloid of the throwing arm, most caudal point on medial/lateral epicondyle of the throwing arm, most dorsal point on the acromioclavicular joint of right/left shoulder, anterior/posterior surfaces of the right (left) shoulder overlying the glenohumeral joint, right/left greater trochanter, processus spinosus of the 7th cervival vertebra, processus spinosus of the 8th thoracic vertebra, deepest point of incisura jugularis, processus xiphoideus, most caudal point on the sternum. Two markers were attached on the ball symmetrically, as well. In an indoor testing facility, after comfortable warm-up subjects were requested to throw an official ball (diameter: 0.21m, mass: 0.38kg) aiming at the target 8m ahead until three successful performances were recorded. Three-dimensional motion analysis system (Vicon-MXB, Oxford Metrics Inc., ten cameras, 250Hz) was used to record the locations of the reflective markers.
The best throwing trial evaluated by the subject himself and the coach was selected for analysis. The location data of the markers were smoothed with the fourth-order zero-lag digital filter of the Butterworth type with optimal frequencies (Winter, 1990). Angular displacement of shoulder and elbow joint was calculated with the projecting method (Sakurai et al., 1993). Kinetic values (joint force and torque) were calculated for the elbow joint using the kinematic data, body segment parameters for primary school children (Yokoi et al., 1986), and inverse dynamics procedures (Feltner and Depena, 1986; Fleisig et al., 1995).

RESULTS: A typical example of changes of the elbow joint torques in varus/valgus, extension/flexion and pronation/supination directions were shown in Figure1. Pronation/supination torque fluctuated in the smallest range, and the varus/valgus torque was in the largest. The peak varus torque (17.4 ± 1.9Nm) was the largest among all the elbow joint torques (Table 1), and occurred just before ball release as is the case in baseball pitching.

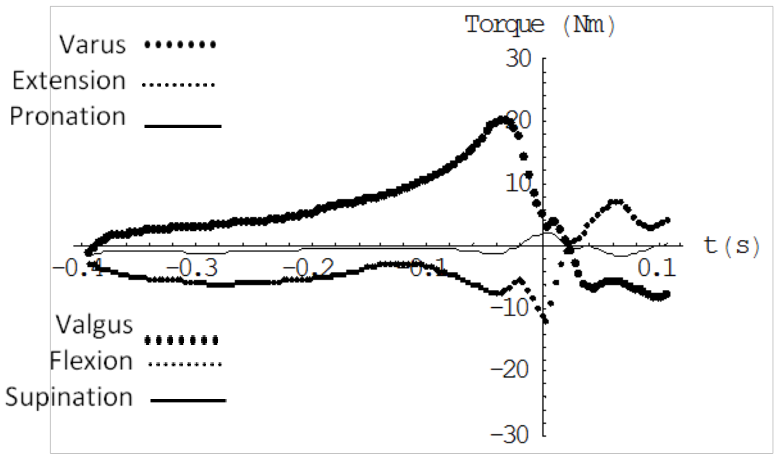

Figure 1. Changes of joint torques at elbow. (The instant of ball release is assigned to t=0.0s)

Table 1. Values of peak torques (Nm) of the elbow joint

	Varus/Valgus Max	Varus/Valgus Min	extension/flexion Max	extension/flexion Min	pronation/supination Max	pronation/supination Min
Torque	17.4 (1.9)	-5.2 (2.3)	4.4 (1.6)	-7.3 (2.7)	1.5 (0.5)	-1.1 (0.3)

Average and (S.D.)

DISCUSSION: For baseball pitching, the load of the ulnar collateral ligament (UCL) due to the varus torque may result in UCL injury (Fleisig et al., 1995). Morrey & An (1983) showed in vitro that when the elbow was flexed 90° and a valgus load was applied to the elbow by the forearm, the UCL generated 54% of the varus torque for resisting valgus motion. While the elbow was extended (0°) , the value was 31%. When the shoulder turned to internal rotation from the maximum external rotation (Figure 2), the peak varus torque (17.4Nm) was applied to the forearm at the elbow joint. At this instant the angle of elbow joint was -64.8° (Figure 2). Assuming the load of UCL from the varus torque is linearly proportional to the elbow joint angle, UCL had generated 8.2Nm when throwing dodgeball. The value (8.2Nm) is clearly smaller than the value in the case of adult baseball pitchers (34.6Nm, Fleisig et al., 1995). However, the force load in UCL is in inverse proportion to the lever arm length of UCL. The lever arm length of children is certainly shorter than for adult. Move over the limit load of UCL rapture for the children could be far smaller than that for adult, which was reported as 32.1Nm by Morrey & An (1983). Varus torque can lead to injuries of muscle or ulnar nerve too (Atwater, A. E., 1979; Fleisig et al., 1995).

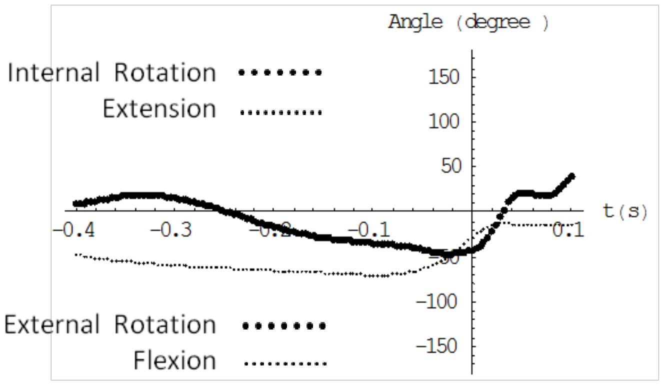

Figure 2. Angular displacement for shoulder Internal/External Rotation and elbow Extension/flexion.

Varus torque can also lead to lateral elbow compressive injury. Morrey & An (1983) reported that in vitro when the elbow was flexed 90°, 33% of the varus torque was caused by the joint articulation. When the elbow was extended (0°), the value was 31%. Assuming the load of joint articulation due to the varus torque is linearly proportional to the elbow joint angle, the torque applied by joint articulation was 5.6Nm for dodgeball throwing motion of the children. Again, the lever arm length is shorter and the strength of bone, cartilage, and soft tissues is weak. Therefore, the kinetic stress in dodgeball throwing could be dangerous for elbow

injuries.
For young baseball player, there are several rules or regulation to prevent shoulder and elbow joint injuries. For example, The quantity of pitching is limited in one game, or the days for rest after games in a certain leagues or an association in Japan. However, for the dodgeball player, there is no any limitation at this stage (Wang et al., 2008). It's necessary for young dodgeball players to set a limit on the throwing number of times of ball throwing similar to young baseball players. In dodgeball games, there are several players with side hand arm motion with extended elbow. For young baseball players, curve pitches have been restricted because mechanical stress could be larger in curve throwing motion. It's also possible for dodgeball players that the kinetic stress is different in some different throwing style compared to classical overhand style. So, it's necessary to analyze other style throwing motion for the young dodgeball players.

CONCLUSION: The changing patterns of elbow joint torques in dodgeball throwing motion of children were similar to those for adult baseball pitchers. It is possible for young dodgeball players to suffer the acute and chronic injuries in elbow joint. For the healthy growth of children, it's necessary to set some rules or regulation to limit the number of times of throwing and to execute further biomechanical studies.

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SHOULDER STABILITY TRAINING AND SHOULDER AILMENTS IN HIGH SCHOOL SWIMMERS

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The purpose of this study was to explore how specific shoulder stabilizing exercises influenced the shoulder strength balance, incidence of practice time lost due to shoulder ailments and swim performance. Two US high school varsity swim teams participated in the study. The treatment team (N = 59) performed thrice weekly 20-minute dryland activities to improve shoulder and scapular stability, whereas the control team (N = 68) did not explicitly train their athletes in this manner. In addition to the shoulder ailment incidence and lost practice time, we also monitored changes in athletic performance. The results indicate that shoulder-stabilizing exercises reduce the incidence and duration of shoulder ailments in swimmers, without being detrimental to swim performance.

KEYWORDS: competitive swimming, high school athletics, injury risk

INTRODUCTION: Swimming is a sports enjoyed by millions throughout the world, both as an activity to increase fitness and as a competitive endeavor. Though the injury rates in swimming is low (2.2 injuries per 1000 swimming occasions) among all swimmers (McFarland & Wasik, 1996), competitive athletes have a high incidence of shoulder ailments, with over 25% of competitive swimmers stating that shoulder pain has their hindered training and performance (McMaster & Troup, 1993).

The high stroke volume in competitive swimming may be one reason for the elevated incidence of shoulder ailments. Swimmers take approximately 2500 strokes per day (Pink and Tibone, 2000), more than 10 times the repetitions that a typical overhead athlete takes. A second theory identifies muscle imbalance, inflexibility, and asymmetry as the cause of increased shoulder ailments in swimmers (Shapiro, 2001), with Jobe and Pink (1993) stating that musculoskeletal instability is at the core of most shoulder injuries. Musculoskeletal instability can result from an asymmetric strength profile, and research suggests that an increased ratio between external and internal rotational strength ratio (ER:IR) may be indicative of shoulder ailments (e.g. Bak and Magnusson, 1997). Specifically, they noted the functional strength ratio of eccentric external rotation to concentric internal rotation (ER_{ecc}:IR_{conc}) were significantly higher in swimmers with symptomatic shoulders.

As the ER:IR ratio is thought to be indicative of shoulder ailments, a specific shoulder stabilizing program that strengthens the internal and external rotators may decrease the incidence of shoulder injuries. Thus, the objective of this work was to explore how introducing specific stabilizing exercises affected the shoulder ER_{ecc}:IR_{conc} ratio and incidence of shoulder ailments over the course of a high school swim season.

METHODS: Two US high school swim team participated in this study (Table 1). The treatment team (N = 59) engaged in thrice weekly 20-minutes exercises to develop shoulder and scapular stability, whereas the control team (N = 68) did not explicitly include these stability exercises, unless directed by a trainer or physician as part of a rehabilitation or treatment program. Participation in the study included parental consent and subject assent prior to the start of the high school season, as well as participation in all assessments. The study protocol was approved by the University's Institution Review Board.

These teams were chosen because they had a similar number of practices per week (8 practices in 6 days with 1 day off) and swim practice hours (approximately 12 hours weekly). Both teams also incorporated as a part of their program 2 hours dryland activities, which included strength training 2 days per week.

Table 1. Subject demographics. Data reported as mean (SD).

	Treatment Team (N = 59)		Control Team (N = 68)	
	Men	Women	Men	Women
Number of Participants	27	32	33	35
BMI (kg/m^2)	16.2 (1.9)#	21.9 (2.6)#	15.1 (2.3)#	22.3 (2.0)#
Age at Start of Study (years)	16.4 (0.7)	15.9 (1.0)	16.7 (0.9)	16.1 (1.0)
Swimming Experience (years)	6.4 (2.1)	6.7 (1.9)	6.0 (2.3)	7.1 (2.3)
Number of Athletes with Prior Incidence of Shoulder Ailments that Limited Swim Participation (% of team)	8 (25.9%)	11 (34.4%)	8 (24.2%)	10 (28.6%)
Number of Athletes Experience Shoulder Ailments at Start of Season	0	1	0	0

= p< 0.05 between men and women within teams

At the start of the high school swim season, we collected the participants' demographic information (Table 1) and provided a survey to collect their swimming and shoulder injury history. For the shoulder injury history questions, subjects were to explain any non-contact shoulder injuries that limited swim practice (e.g., requiring the subject to refrain from specific swim activities) and to note outcome of the injury (e.g., surgery, rehabilitated, currently still injured, etc.).

On day 2 of the season and 3 weeks prior season's conclusion, participants completed a pre-post swimming assessment of 10x100m on a 2:20 interval. The average 100m swim time was determined for each participant.

The day following the swim assessments, we determined the ER:IR ratios. We followed the protocol by Bak and Magnusson (1997) and used a KinCom dynamometer (Chattecx Corp., Chattanooga Group, Chattanooga, TN, USA). For the testing, we positioned the shoulder at 80° of abduction and 20° of forward flexion and the elbow at 90° of flexion. Subjects had eight warm-up submaximal efforts at 30°/second prior to testing. In the testing, participants underwent maximal efforts of concentric and eccentric internal rotation strength followed by external rotation strength. There was a self-selected rest period of 30-60 seconds between bouts, and testing continued until participants were unable to demonstrate improved strength scores. We normalized the ER:IR ratios by body mass, and the participant's right and left ratios were averaged.

Throughout the season, the coaches of the participating teams emailed the author the injury report provided from the high school, which listed injuries and contraindication for practice. Any contraindications or limitation for practice as a result of shoulder pain were verified by the coach to explain if the athlete refrained from stated activities.

The shoulder exercises (Table 2) addressed the three important areas in shoulder joint's stability, and followed the protocol set by USA Swimming (Rodeo, 2002). The program sought to

Table 2. Treatment team exercises.

Shoulder Stabilizing Exercises		
Rotator Cuff Exercises	**Actions**	**Main Muscles**
External Rotation with Resistance Band	With resistance band in circle, hands are placed in the circle and press the laterally with elbows maintaining close contact with trunk.	Deltoid Infraspinatus Teres Minor
Light Weight Scaption (Straight Arm Lifts)	Using 1-3 lb weights per hand, straight arms lift the weight obliquely at 45° angle.	Supraspinatus
Ball on the Wall	With one arm extended and a tennis ball against the wal, shoulder blades squeeze together and roll the ball in small circles in a counter-clockwise and clockwise motion.	Supraspinatus Infraspinatus Teres Minor Subscapularis
Shoulder Exercises	**Actions**	**Main Muscles**
Resistance Band Low Row	With resistance band connected to stationary object, squeeze shoulder blades together, palms facing upward, bring hands towards body leading with the elbows.	Latissimus Dorsi Rhomboids Posterior Deltoid Trapezius
Prone Reverse Fly	Lying face down on the floor, arms make a 'T' with body, and squeezing shoulder blades together, arms lift off of the ground.	Rhomboids
Push Up Plus	With a regular pushup, at the top of a pushup, continue to push such that shoulders rotate and center of back is farther away from the floor than shoulders.	Pectoral Anterior Deltoid Triceps Latissimus Dorsi Rotator Cuff Muscles
Core Exercises	**Actions**	**Main Muscles**
Dead Bug	Lying supine with a neutral back, hold abdominals tight and light flutter kick, arms can also light alternate overhead.	Rectus Abdominis Transverse Abdominis
Quadraped	On hands and knees, hold a neutral back and slight press one leg behind and the opposite arm forward. Alternate in a slow, controlled motion.	Transverse Abdominis
General Stretches	**Actions**	**Main Muscles**
Hamstrings	Lying supine, with a straight leg, bring leg towards chest.	Hamstrings
Upper Back	With palms together and straight arms at shoulder height, press hands forward, rounding the back,	Trapezius Rhomboids
Neck	With one arm placed behind the back, lightly press the head to the opposite shoulder. Eyes will look forward for one stretch and then eyes will look at a 45° downward for a second stretch.	Trapezius

strengthen the: (1) rotator cuff muscles, (2) scapular stabilizing muscles, and (3) muscles of the low back, abdominal, and pelvis
We used SPSS 15.0 (SPSS Inc, Chicago, IL, USA) for the statistical analysis. The average swim times, shoulder ailments and days of limited swim practice were calculated. ANOVA used to analyze pre-post assessments and ER_{ecc}:IR_{conc}, in addition to incidence and duration of the shoulder ailment.

RESULTS: Throughout the course of the season, the control team men (p=0.045) and women (p=0.048) experienced significant higher loss of practice due to injury compared to the treatment team men and women (Table 3). These injuries, on average, resulted in significantly more limited swim practices (p = 0.039) when compared to the treatment team.
In exploring the changes in swim performance between the two teams, the female treatment group showing significant improved (p = 0.044) over the course of the season. This improvement in swim performance is significant, as members of the treatment team reduced their time in the water swimming by one hour to accommodate the shoulder stabilizing exercises during practice.
At the pre-assessment, the functional ER_{ecc}:IR_{conc} strength ratio between team was initial similar between genders and teams.

DISCUSSION: The purpose of this study was to explore how shoulder stability exercises affected the functional ER_{ecc}:IR_{conc} strength ratio and shoulder injury during the course of a high school swim team. There were several significant findings. First, we found that adding shoulder stabilizing exercises does decrease the incidence of shoulder ailments. Second, we noted that shoulder stabilizing exercises can significantly reduced the ER_{ecc}:IR_{conc} strength ratio in both men and women. Finally, we observed that swimmers who routinely performed the shoulder stabilizing exercises experienced a quicker return to practice post-complaint. This data suggest that incorporating shoulder-stabilizing exercises in swim practice may benefit the overall shoulder joint health of the swimmer.
Because of the nature of swimming, swimmers tend to have an increase in internal rotation strength development, and the imbalance between internal and external rotation strength may play a role in higher rate shoulder ailments in swimmers (Bak and Magnusson, 1997). Though the shoulder-stabilizing exercises did not explicitly focus on increasing the shoulder external rotation strength, it does provide a means for emphasizing shoulder health during practice, and it did lower the functional ER_{ecc}:IR_{conc} strength ratio.
From the pre to post-assessment, the treatment teams ER_{ecc}:IR_{conc} was significantly lower for the men and women. The treatment team also had a decrease in shoulder ailments incidence. This could be treatment team's lowered ER_{ecc}:IR_{conc} ration. Moreover, it could be that the swimmers decreased instability through strengthening the smaller supporting

Table 4: Pre-Post assessment results and incidence of injury. All data reported as mean (SD).

	Treatment Team (N = 59)		Control Team (N = 68)	
	Men (N = 27)	Women (N = 32)	Men (N = 33)	Women (N = 35)
Pre-Average Timed 100m Freestyle Repeats (10x100m) (sec)	74.0 (4.9)	80.6 (4.1)†	72.9 (4.5)	79.1 (7.0)
Post-Average Timed 100m Freestyle Repeats (10x100m) (sec)	71.2 (5.1)	74.3 (3.9)†	70.8 (4.4)	74.1 (5.1)
Pre ER_{ecc}:IR_{con} Ratio	1.01 (0.13)†	1.03 (0.13)†	1.01 (0.14)	1.01 (0.13)
Pre ER_{ecc}:IR_{con} Ratio	0.91 (0.09)†*	0.93 (0.10)†	0.99 (0.15)*	1.00 (0.13)
Athletes with shoulder ailments limiting swim participation during season (% of team)	2 (7.4%)	3 (9.4%)	5 (15.2%)	5 (14.3%)
Athletes with shoulder ailments (Number of new ailments / Number of reoccurring previous ailment)	0 / 2	1 / 2	2 / 3	3 / 3
Average number of limited swim participation days per shoulder complaint	2.4 (0.3)*	2.0 (0.3)*	3.1 (0.5)*	3.1 (0.4)*

* p < 0.05 between treatment and control teams by gender
† p < 0.05 between pre- and post assessment within group

muscles of the rotator cuff (McMaster, 1999). Further research is needed to elucidate the shoulder adaptations to these exercises to understand their effectiveness in limiting shoulder maladies in swimmers.

CONCLUSION: This research suggests that incorporating specific shoulder exercises into a swim program may be one method to decrease the incidence and duration of swimmer's shoulder ailments. There was a significant difference noted between the athletes that experienced shoulder ailments in the two teams. The treatment team with the added shoulder stability exercises had only 1 athlete develop a new shoulder ailment, whereas the control team had 5 new cases. This is an important finding, and we conclude that incorporating shoulder stability exercises early and throughout an athlete's career may decrease the risk of developing shoulder ailments and reduce practice time lost, which may result in increased longevity in swimming.

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INFLUENCE OF BODY MASS INDEX ON ROWING KINEMATICS

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Rowing meets the criteria for weight loss as defined by the American College of Sport Medicine. Little research exists on the influence of body shape on movement kinematics. Even if rowing is a non weight bearing exercise, the body shape may have an influence on rowing motion and joint reactions. This study investigated rowing movement kinematics between normal weight and overweight subjects. Differences were found for hip abduction and adduction angles. This knowledge can help to understand the influence of body weight and body shape on movement kinematics and can help to avoid overloading the joints.

KEYWORDS: Rowing, motion analysis, body shape

INTRODUCTION: Rowing combines endurance exercise as well as resistance training and has a favorable influence on health and the prevention of diseases. Studies have shown that rowing reduces the risks of falls, limb disability and coronary artery diseases (Yoshiga & Higuchi, 2002). Rowing also has a positive influence on the risk of developing type 2 diabetes, blood pressure, the ability to oxidize long chain fatty acids, the metabolic rate, glycogenic control, lipoprotein profile, hypertension and increases the fat-free mass (Sanada et al., 2009). In order to lose weight rowing can be a beneficial sport for all individuals because it is non-weight bearing and may not produce excess joint forces. Rowing meets all criteria for weight loss as defined by the American College of Sport Medicine (Sanada et al., 2009). However, the body shape may have an influence on rowing motion. Current studies show that body shape influences human gait. Changes in movement angles and increased joint forces in walking are well documented (cf. Browning et al., 2007; Lai et al., 2008). The aim of this study was to compare and analyze the rowing motion of over and normal weight individuals. Assessing kinematics during rowing in over and normal weight individuals may lead to manipulation in ergometer design to decrease the risk of injury.

METHODS: The World Health Organization (WHO) recommended classification of body shapes with the help of the Body Mass Index (BMI). The classifications in Table 1 were adopted in this study.
Ten overweight (five men, five women) and ten normal weight volunteers (five men, five women) with no or little previous rowing experience were recruited. Table 2 shows the characteristic data of the subjects. Exclusion criteria for participants included any past or current neurological or cardiovascular illness and any pain, which might affect their rowing motion. Prior to the investigation, all subjects gave informed consent according to the human subject ethics approval of the Institutional Review Board.

Table 1. Classification for body shapes

Class	BMI (kg/m²)
Normal weight	18.5-24.9
Over weight	25-29.9
Obese	≥ 30

WHO (2004)

Table 2. Means and standard deviations of age, height, weight and BMI of overweight and normal weight subjects

	Normal weight	Over weight
Sample	10 (5 F & 5 M)	10 (5 F & 5 M)
Age (years)	27.4 ± 7.2	24.1 ± 5.9
Height (meter)	1.74 ± 0.08	1.71 ± 0.09
Weight (kg)	66.5 ± 9.2	78.1 ± 9.3
BMI (kg/m²)	21.8 ± 0.5	26.7 ± 0.4

A motion analysis system (Vicon, MX+, Oxford, United Kingdom) was used to collect kinematic data. Body mass, height, and body composition were measured of all subjects by a segmental body composition analyzer (Tanita, BC-418 Pro, Arlington Heights, USA), as well as segmental measurements of the whole body. The subjects were asked to wear tight fitting non-reflective clothes. Reflectors on shoes or clothes were covered by tape. 34 reflective markers where attached to the subject with double-faced adhesive tape (see Figure 1). Also, the rowing ergometer (Concept2, Model E, Morrisville, USA) was equipped with 13 markers (front and back of the ergometer, left and right handle, seat, upper footrest, lower footrest, footrest heel and middle seat). After an introduction on the rowing technique, the subjects performed a short warm-up to practice the rowing technique at a desired stroke rate (23-25 strokes per minute). Each subject rowed at three different resistance levels (3, 5 and 7) for two minutes each and had a break of two minutes between the trials. The second minute of the rowing interval was captured with a frequency of 200Hz. The Man-Model Dynamicus program (Alaska 6.01, Institute of Mechatronics, Chemnitz, Germany) was used to reconstruct the motion and calculate velocities and joint angles.

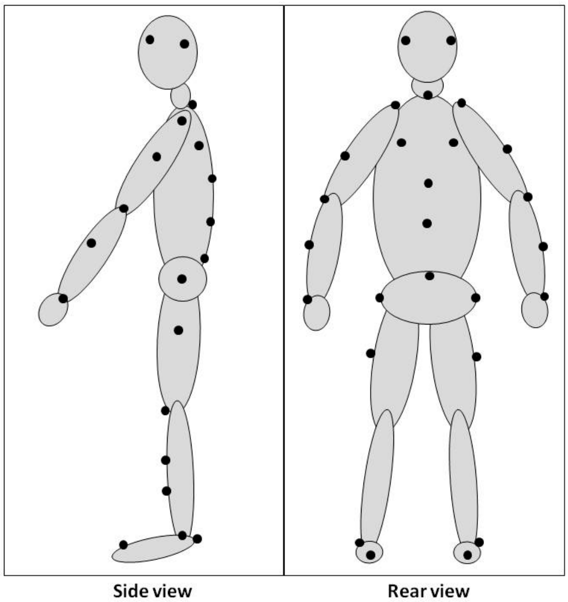

Figure 1. Position of attached markers on the subjects (left: sagittal plane, right: frontal plane)

Minimum and maximum hip, knee, ankle flexion as well as hip adduction were investigated. To make sure that the data were not influenced by acceleration as well as deceleration phases, rowing strokes in the middle of the rowing trial were used. Data were averaged over 10 rowing strokes and segment side. The stroke rate was not normalized because the point of interest was just range of motion. A Kolmogorov-Smirnov-Test was used to test the homogeneity of the data. Additionally, an independent samples t-test was performed to examine the effect of BMI on the aforementioned joint ranges of motion. The significance level was set at $p \leq 0.05$.

RESULTS: Two normal weight subjects (one female and one male) could not be included in the results because of missing data points. All data were homogenous ($p \geq 0,343$). Significant differences were found with respect to hip adduction (p=0,003) and abduction (p=0,045) between normal weight and overweight subjects. All other joint angles were not statistically different between the weight groups.

Table 3. Mean and standard deviation hip, knee and ankle flexion and hip adduction

	BMI_Group	N	Mean	Std. Deviation	t	df	p
hip flexion / extension	normal weight	8	52.9 / 105.7	13.5 / 8.9	-0.456 / -0.031	16 /16	0.654 / 0.975
	over weight	10	56.1 / 106	15.8 / 17.1			
hip adduction / abduction	normal weight	8	-2.0/ 2.0	1.8 / 0.9	-3.452 / -2.175		0.003* / 0.045*
	over weight	10	0.3 / 4.9	1.0 / 3.7			
knee flexion / extension	normal weight	8	-138.0 / -21.1	5.7 / 9.9	-1.318 / -1.150		0.206 / 0.267
	over weight	10	-132.9 / -16.8	9.6 / 5.9			
ankle flexion / extension	normal weight	8	-9.0 / 37.9	9.0 / 5.0	1.001 / - 0.499		0.332 / 0.625
	over weight	10	-13.0 / 39.9	8.5 / 8.6			

* Significant difference between normal and overweight subjects p<0.05.

DISCUSSION: The results revealed that the body shape does influence kinematic variables of rowing. The hip adduction and abduction were significantly different for the two weight groups indicating different movement strategies. More specifically, the overweight subjects demonstrated higher hip abduction and less hip adduction during the catch phase, possibly due to their body shape. No differences for the hip flexion and extension angles were found. The kinematic differences between the normal and overweight groups may provide insight into injury prevention.

Low back pain is a common injury for elite rowers (McNally & Seiler, 2005). The ability to produce a high flexion angle in the hip during the catch phase is related to the likelihood of developing low back pain in rowers (Soper et al., 2004). The increased risk for low back in rowers way be linked to kinematics, overtraining or skill level. It is known that in particular the hip flexion angle is closely related to skill level in rowing (Soper & Hume, 2004). Even though the subjects in this study had no or little previous experience in rowing on a rowing ergometer, their ability to perform the rowing technique might have influenced the results. However, the similarities in hip flexion angle between the normal and overweight subjects suggest that rowing may present similar risks for the groups.

The similar knee and hip flexion angles in the overweight subjects may indicate that the increased abduction angles were a compensation for the likely greater abdominal mass. However, fat distribution was currently not measured, thus reasons for the differing hip ab/adduction angles cannot be certain. Most elite rowers have a BMI over 25 due to their increased muscle mass (Skinner et al., 2010; Izquierdo-Gabarren et al., 2010). However, the participants in this study were volunteers with little or no previous experience. Therefore, an increased BMI due to increased muscle mass is unlikely. This presents a limitation in using BMI to assess body composition. These results indicate that there may be more differences in rowing technique for individuals with larger BMI (over 30) or higher percent body fat. Practical application to the differences in rowing kinematics between body sizes may involve changes to ergometer design.

In elite rowing sports the boat setups are variable. However, rowing ergometers are non-adjustable. The results of this study suggest that manipulation of the rowing ergometer (i.e. adjustable footrests, wider seats etc.) for different body shapes may be useful to remediate any kinematic variations and increase comfort during rowing. However, further studies (different skill levels, greater BMI variability etc.) are needed to gain insight into the differences in rowing kinematics between varying body types and how equipment manipulations effect these changes.

CONCLUSION: This study investigated rowing movement kinematics between normal weight and overweight subjects. Differences were found only for hip abduction and adduction angles. Manipulation of ergometer design may help limit these positions. The goal for future studies should be to investigate the kinematics between more widely varying body sizes (normal weight, overweight and obese) as well as the influence of different skill level and ergometer design on these groups.

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SPATIAL－TEMPORAL ANALYSIS OF BUTTERFLY STROKE PATTERN

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KEYWORDS: coordination, butterfly stroke, percent of cycle.

INTRODUCTION: The butterfly stroke is a unique swimming style and has been considered as the most difficult swimming style to perform because of the synchronizing characteristics of the arm and leg movements. The aim of this study was to examine the different phases of arm coordination patterns in the butterfly stroke between two levels of swimmers who were specialized in the butterfly stroke at three race paces (50M, 100M and 200M).

METHOD: Eight Taiwanese elite swimmers and eight division B college swimmers participated in the study. Two underwater high-speed cameras (200 Hz) were set in the transaction and sagittal plane under the water and synchronized to capture the swimming movement. The kinematics data were digitized and calculated with the Kwon 3D software. The catch phase is made from the fingers enter water to sweep outside of the widest point. The pull phase occurred at the end of catch phase to the minimum elbow joint angle under the chest. The push phase was from the end of pull phase to the fingers exist water.

RESULTS: The results show that there were significant differences between two levels, $F(1,14)=21.336$, $p<.05$ and three difference velocity in each group, $F(2,14)=5.056$、$F(2,14)=6.720$. In elite swimmers, the catch and pull phase had a significant difference by race paces, $F(2,14)=3.901$; $F(2,14)=4.386$, $p<.05$. The non-elite swimmers had a significant difference by race paces in pull phase and recovery, $F(2,14)=0.412$; $F(2,14)=9.705$, $p<.05$.

DISCUSSION: The elite butterfly swimmers demonstrated a decreasing percentage of duration for the catch phase over the increasing swimming velocity. In other word, by increasing the velocity, the elite butterfly swimmers had the trend potential of shorten the percentage of catch phase. The pull phase changed with velocity in two groups. Non-elite swimmers as the velocity increase up in recovery phase at the same time.

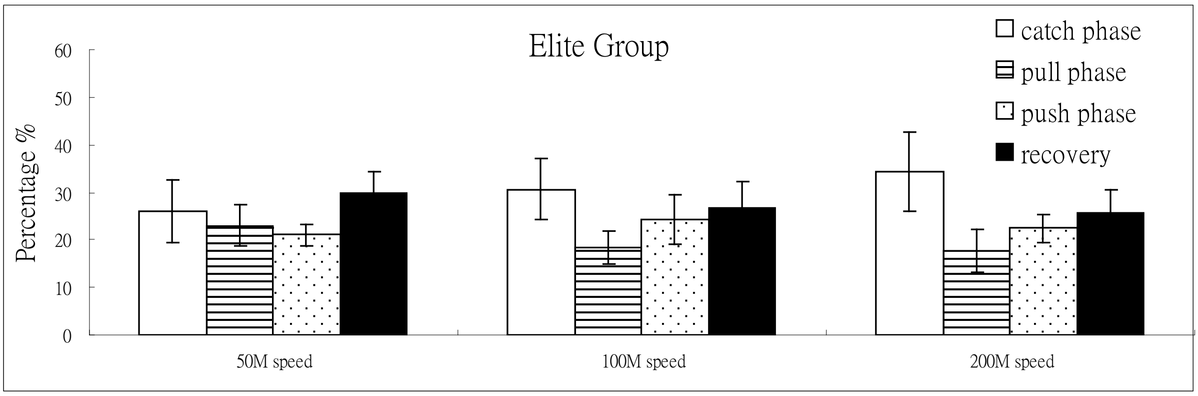

Figure 1. The percentage of each arm phases of butterfly stroke in elite group

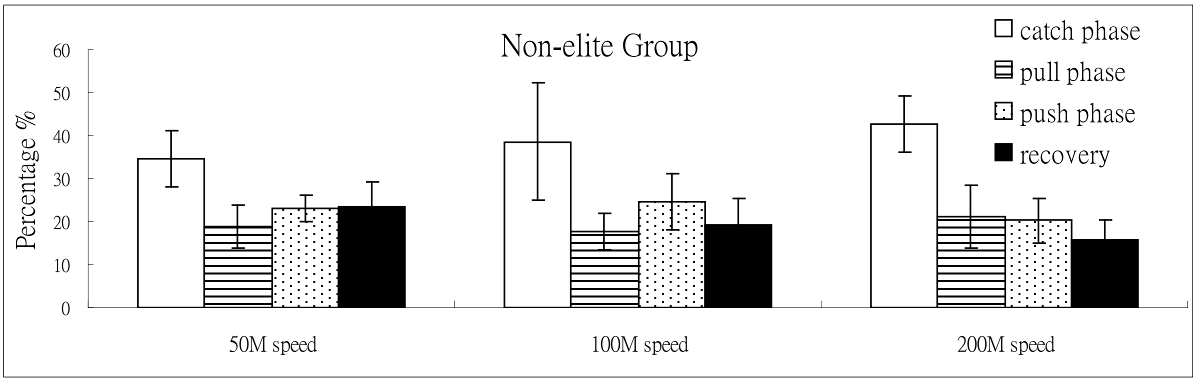

Figure 2. The percentage of each arm phases of butterfly stroke in non-elite group

CONCLUSION: In catch phase, there is an obvious descending trend with velocity in two groups. However, even in elite swimmers can hardly find an identical pattern in each race paces. The percentage of arm stroke imply the coordination of the butterfly stroke, the consistency should include in each race paces.

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GROUND REACTION FORCES OF VARIATIONS OF PLYOMETRIC EXERCISES ON HARD SURFACES, PADDED SURFACES AND IN WATER

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Subjects performed drop jumps from 46 cm, a single leg jump, counter movement jump, and squat jump on a hard surface, wrestling mat and in water. Ground reaction force data obtained via a force platform were used to determine the time to takeoff, takeoff peak ground reaction force, power, jump height, and landing peak ground reaction force. A one way repeated measures ANOVA demonstrated differences between plyometic exercises assessed for all of the variables assessed ($P \leq 0.05$), with post hoc analysis demonstrating the specific differences. Results indicate that the hard surface and mat conditions were similar for almost all of the plyometric exercises assessed for most outcome variables whereas the plyometric exercises performed in water were different than those performed on the hard surface or mat in most cases.

KEYWORDS: stretch shortening cycle, intensity, stress, progression, aquatic, compliance

INTRODUCTION: Recommendations have been made to perform plyometric exercises on surfaces that are neither too hard nor soft, since these surfaces are thought to increase injury potential or prolong the amortization phase, respectively (Potach & Chu, 2004). In an attempt to evaluate the effects of surface type on plyometric training, studies have compared plyometric training in water versus a control group (Martel et al., 2005), plyometric training on land and in water (Miller et al., 2002; Robinson et al., 2004; Stemm et al., 2007), and training on grass versus sand surfaces (Impellizzeri et al., 2008). Some evidence indicates no differences in the outcome variables assessed between land and aquatic plyometrics (Robinson et al., 2004; Stemm et al., 2007). Other studies show increases in countermovement jump performance on grass compared to sand (Impellizzeri, et al., 2008), and less soreness in the aquatic conditions (Impellizzeri, et al., 2008; Robinsion et al., 2004). At this point, little information exists with respect to the acute effects of performing plyometrics on varying surfaces.

Studies have examined acute biomechanics effects of plyometric exercises (Crowther et al., 2007; Gaitsis et al., 2008), isokinetic force (Miyama & Nosaka, 2004), muscle soreness (Miyama & Nosaka, 2004), ground reaction forces (GRF) (Gaitsis et al., 2008; Miyama & Nosaka, 2004) and power and take off velocity (Gaitsis et al., 2008; Miyama & Nosaka, 2004). These studies have compared plyometrics performed on the ground versus a mini-trampoline (Crowther et al., 2007) and on hard versus sand surfaces (Gaitsis et al., 2008; Miyama & Nosaka, 2004). Research indicates that force, power, and take off velocity are higher on rigid compared to compliant surfaces during the take off phase of the plyometric exercise (Gaitsis et al., 2008). Ground reaction forces have been shown to be higher on compliant surfaces during the landing phase of some plyometric exercises, potentially due to stiffer landings as a result of lower levels of lower body joint flexion on compliant surfaces such as sand (Crowther et al., 2007). Thus, compliant surfaces may not necessarily reduce plyometric exercise intensity.

Studies assessing plyometric exercises on a variety of landing surfaces have compared a limited number of exercises including the depth jump (Crowther et al., 2007; Miyama & Nosaka, 2004), countermovement jump (Crowther et al., 2007), and squat jumps (Gaitsis et al., 2008). Typically these only compared two exercises and two landing surface variations (Crowther et al., 2007; Gaitsis et al., 2008; Miyama & Nosaka, 2004) and did not examine the takeoff phase of the exercises. Other research compared the intensity of a variety of plyometric exercises, but did not investigate the effect of different types of landing surfaces (Jensen and Ebben, 2007). Therefore, the purpose of the present study was to evaluate the kinetic characteristics of a variety of plyometrics performed on a hard surface, wrestling mat, and in water.

METHODS: Fifteen men and women (mean ± SD; age = 21.2 ± 2.2 years, height = 170.3 ± 6.5 cm; body mass = 68.81 ± 12.15 kg) who were familiar with and used the studied exercises served as subjects in this study. Subjects provided informed consent prior to participating in the study, which was approved by the institutional review board.
Prior to the test subjects warmed up with at least 3 minutes of low intensity work on a cycle ergometer and stretched for approximately 12 seconds using one exercise for each major muscle group. Subjects then rested at least 5 minutes rest prior to beginning the test plyometric exercises. The test plyometric exercises included a drop jump (DJ) from 46 cm, single leg jump using the right leg (SLJ), counter movement jump (CMJ), and a squat jump (SJ), each performed in a randomly assigned order. A one minute rest interval was maintained between test plyometrics.
The plyometric exercises were performed on a hard surface, mat, and in water. The hard surface condition included performing the plyometric exercises on a 2cm thick aluminum plate (76 X 102 cm) bolted directly to a force platform (OR6-5-2000, AMTI, Watertown, MA, USA). Attachment of the plate resulted in a natural frequency of not less than 142 Hz, within limits recommended for this system. For the mat condition, a section of 5 cm thick closed cell wrestling mat was attached to the surface of the force platform. The water condition included performing the plyometric exercises on a force platform (OR6-WP-2000, AMTI, Watertown, MA, USA) which was placed on the pool bottom at a depth of 140 cm.
Ground Reaction Force data were collected at 1000 Hz, real time displayed and saved with the use of computer software (NetForce 2.0, AMTI, Watertown, MA, USA) for later analysis. Takeoff peak GRF, time to takeoff, power, jump height, and landing peak GRF were calculated from methods previously used (Bauer et al., 2001; Jensen & Ebben, 2007).
Data were analyzed using a repeated measures ANOVA to test main effects for takeoff peak GRF, time to takeoff, power, jump height, and landing peak GRF. Bonferroni adjusted post hoc analyses were used to assess the specific differences between the plyometric exercises. The *a priori* alpha level was set at $P \leq 0.05$. Statistical power (d) and effect size (η_p^2) were determined and all data expressed as means ± SD.

RESULTS: Significant main effects representing differences between plyometric conditions were found for the CMJ for takeoff peak GRF ($P = 0.034$, $\eta_p^2 = 0.23$, $d = 0.65$), time to takeoff ($P \leq 0.001$, $\eta_p^2 = 0.41$, $d = 0.96$) power ($P \leq 0.001$, $\eta_p^2 = 0.72$, $d = 1.00$), jump height ($P \leq 0.001$, $\eta_p^2 = 0.72$, $d = 1.00$), and landing peak GRF ($P \leq 0.001$, $\eta_p^2 = 0.66$, $d = 1.00$).
Significant main effects representing differences between plyometric conditions were found for the SLJ for takeoff peak GRF ($P=0.025$, $\eta_p^2 = 0.76$, $d = 1.00$), power ($P \leq 0.001$, $\eta_p^2 = 0.97$, $d = 1.00$), jump height ($P \leq 0.001$, $\eta_p^2 = 0.97$, $d = 1.00$), and landing peak GRF ($P \leq 0.001$, $\eta_p^2 = 0.43$, $d = 0.98$), but not for time to takeoff ($P = 0.12$).
Significant main effects representing differences between plyometric conditions were found for the SJ for takeoff peak GRF ($P \leq 0.025$, $\eta_p^2 = 0.25$, $d = 0.70$), time to takeoff ($P \leq 0.001$, $\eta_p^2 = 0.56$, $d = 0.99$), power ($P \leq 0.001$, $\eta_p^2 = 0.86$, $d = 1.00$), jump height ($P \leq 0.001$, $\eta_p^2 = 0.86$, $d = 1.00$), and landing peak GRF ($P \leq 0.001$, $\eta_p^2 = 0.57$, $d = 0.99$).
Significant main effects representing differences between plyometric conditions were found for the DJ for takeoff peak GRF ($P = 0.001$, $\eta_p^2 = 0.54$, $d = 0.99$), power ($P \leq 0.001$, $\eta_p^2 =$

0.88, d = 1.00), jump height ($P \leq 0.001$, $\eta_p^2 = 0.88$, d = 1.00), and landing peak ground reaction force ($P \leq 0.001$, $\eta_p^2 = 0.71$, d = 1.00), but not for time to takeoff (P = 0.93).
Results of the post hoc analysis demonstrating specific differences between plyometric conditions are shown in Tables1-5.

Table 1. Time to takeoff in seconds (Mean ± SD) for each plyometric exercise

	CMJ*	SLJ	SJ**	DJ
Land	0.66 ± 0.10	0.75 ± 0.26	0.48 ± 0.12	0.29 ± 0.09
Mat	0.60 ± 0.17	0.70 ± 0.19	0.48 ± 0.07	0.29 ± 0.09
Water	0.87 ± 0.31	0.86 ± 0.25	0.33 ± 0.04	0.29 ± 0.10

CMJ = countermovement jump, SLJ = single leg jump, SJ = squat jump, DJ = depth jump
*Significant difference between the Land and Water and Mat and Water conditions ($P \leq 0.05$)
** Significant difference between the Land, Mat and Water ($P \leq 0.05$)

Table 2. Peak ground reaction force, minus body mass, in Newtons (mean ± SD) during the takeoff phase of each plyometric exercise

	CMJ*	SLJ**	SJ*	DJ**
Land	921.83 ± 244.38	607.42 ± 230.04	807.51 ± 172.21	1917.81 ± 750.51
Mat	938.31 ± 314.61	612.99 ± 191.44	784.75 ± 205.87	1880.42 ± 615.05
Water	782.35 ± 353.72	264.73 ± 176.47	701.99 ± 245.87	1333.97 ± 357.61

CMJ = countermovement jump, SLJ = single leg jump, SJ = squat jump, DJ = depth jump
*Significant difference between the Land and Water and Mat and Water conditions ($P \leq 0.05$)
** Significant difference between the Land and Water and Mat and Water conditions ($P \leq 0.001$)

Table 3. Power in watts (mean ± SD) for each plyometric exercise

	CMJ*	SLJ*	SJ*	DJ*
Land	30869.16 ± 5837.22	29434.84 ± 5440.38	29674.31 ± 5034.89	30261.4779 ± 5459.03
Mat	30745.62 ± 5467.47	29489.14 ± 5433.48	29666.22 ± 5054.60	30282.93 ± 5436.72
Water	32683.95 ± 5159.68	32187.00 ± 5380.80	31811.22 ± 5083.95	32657.94 ± 5021.38

CMJ = countermovement jump, SLJ = single leg jump, SJ = squat jump, DJ = depth jump
*Significant difference between Land and Water, and Mat and Water conditions ($P \leq 0.001$)

Table 4. Jump height in meters (mean ± SD) for each plyometric exercise.

	CMJ*	SLJ**	SJ*	DJ*
Land	0.35 ± 0.08	0.15 ± 0.05	0.31 ± 0.08	0.29 ± 0.09
Mat	0.33 ± 0.10	0.16 ± 0.06	0.31 ± 0.08	0.29 ± 0.09
Water	0.65 ± 0.17	0.60 ± 0.07	0.66 ± 0.13	0.68 ± 0.18

CMJ = countermovement jump, SLJ = single leg jump, SJ = squat jump, DJ = depth jump
* Significant difference between the Land and Water, and Mat and Water conditions ($P \leq 0.001$)
** Significant difference between the Land and Mat ($P \leq 0.05$) and Land and Water and Mat and Water conditions ($P \leq 0.001$)

Table 5. Landing peak ground reaction force in Newtons (mean ± SD) for each plyometric exercise

	CMJ*	SLJ*	SJ*	DJ**
Land	2614.07 ± 1181.04	1457.82 ± 631.34	2413.16 ± 1681.38	3266.61 ± 1059.36
Mat	3088.75 ± 1492.73	1534.54 ± 593.67	2566.00 ± 1277.61	2796.38 ± 710.97
Water	782.35 ± 353.72	717.62 ± 887.44	789.97 ± 968.87	1546.12 ± 555.71

CMJ = countermovement jump, SLJ = single leg jump, SJ = squat jump, DJ = depth jump
*Significant difference between the Land and Water and Mat and Water conditions ($P \leq 0.001$)
** Significant difference between the Land and Mat ($P \leq 0.01$) and Land and Water and Mat and Water conditions ($P \leq 0.001$)

DISCUSSION: This is the first study to compare the kinetic properties of variety of plyometrics performed on a hard surface, mat, and in water. Results demonstrate that the take off characteristics including time to takeoff do not differ between the hard surface and mat conditions. This finding indicates that this type of padded surface does not appear to prolong the stretch shortening cycle. This finding is consistent with work that demonstrated no difference in the total time to execute the countermovement jump or depth jump when performed on the mini-trampoline or ground (Crowther et al., 2007). Time to take off in the water was longer than the other plyometric conditions in the present study, potentially due to lower landing GRF for depth jumps and a longer eccentric phase for the other plyometric exercises, which may not optimally stimulate the stretch shortening cycle or produce the minimal essential eccentric strain (Ebben et al., 1999). Takeoff GRF have previously been demonstrated to be higher on rigid compared to compliant surfaces (Gaitsis et al., 2008) though no differences were found in the present study between hard surface and mat for any of the exercises assessed. However, in the present study, takeoff peak GRF were lower for plyometric exercises performed in water.

Jump heights and power values were higher in the aquatic compared to the hard surface and mat conditions. Anecdotal observations suggest that jump heights are elevated in water, due to its buoyancy. Additionally, the jump heights and power values are based on calculations using flight time equations which may be falsely elevated due to the buoyancy of the water during the landing phases of the plyometrics. Gaitsis et al. (2008) found that power values were higher on a rigid surface compared to sand, whereas no difference was found between hard surfaces and mats in the present study. Landing peak GRFs were lower for the plyometric exercise performed in water. This finding suggests that aquatic plyometrics are less intense which may lead to less chronic muscle soreness, consistent with previous reports (Miller et al., 2002; Robinson et al., 2004).

CONCLUSION: These data indicate that plyometrics performed on a hard surface and a mat demonstrate similar take off and landing kinetics. Compared to plyometrics performed on hard surfaces and mats, plyometrics performed in water produce lower take off and landing ground reaction forces, slower times to take off, but produce elevated power and jump heights which may be falsely inflated due to flight time based equations.

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THE EFFECT OF WHOLE BODY VIBRATION ON THE DYNAMIC STABILITY OF WOMEN BASKETBALL PLAYERS

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This study investigated the effect whole body vibration (WBV) on the dynamic stability of NCAA Division I women basketball players. Eleven subjects were evaluated in two test conditions including one with and one without WBV. After each condition, subjects were tested for time to stabilization (TTS) on a force platform during bilateral, right leg, and left leg countermovement jumps (CMJ). Results of the statistical analysis revealed no significant difference in TTS between the vibration and non-vibration conditions for the bilateral (p = 0.24) and right leg (p = 0.48) CMJ. A significant difference was found between the conditions demonstrating a shorter TTS in the non-vibration condition for the left leg CMJ (p = 0.04, d = 0.57, η_p^2 = 0.36). Acute WBV has no effect on and in some cases impairs dynamic.

KEYWORDS: Time to stabilization, countermovement jump, landing, postural stability.

INTRODUCTION: Whole body vibration (WBV) has been used for rehabilitation, stimulating bone development, and enhancing physical performance. The acute enhancement of neuromuscular performance after vibration may be due to increased sensitivity of the stretch-reflex (Martin & Park, 1997). The vibration mediated deformation of soft tissues may activate muscle spindles, leading to an enhancement of a stretch-reflex loop. Cardinale & Bosco (2003) suggest that vibration could represent an effective exercise intervention for enhancing neuromuscular performance of athletes by increasing muscle activation via the stretch reflex.

Single and multiple exposures of short-term WBV training have been shown to improve power and strength (Bosco et al., 1999) and vertical jumping ability (Cardinale & Bosco, 2003). The acute application of WBV ranging from 30 seconds (Cormie et al., 2006) to 5 minutes (Bosco, 1999) has resulted in subsequent countermovement jump (CMJ) height increase. Similarly, training studies incorporating WBV ranging from eight weeks (Fagnani et al., 2006) to four months (Torvinen et al., 2002b), have resulted in increase subject strength, CMJ height, and flexibility. On the other hand, some evidence shows no positive acute effect of a four minute WBV session on strength and CMJ height (Torvinen et al., 2002a).

In addition to strength and power performance, the effect of WBV on balance has been studied. Results are equivocal with some studies demonstrating no improvements (Torvinen et al., 2002a; Torvinen et al., 2002b) while others found significant improvement in balance after acute WBV exposure (Moezy et al., 2008; Torvinen et al., 2002c). Previous studies that assessed balance used the Biodex Stability System (BSS) (Biodex Medical Systems, Shirley, NY, USA). The BSS measures the degree and time of tilt about unstable axes. Unfortunately, since foot stance in most sports does not occur on a surface with an unstable rotary axis, the BSS does not measure athletic activity (Wickstrom et al., 2005) and therefore may lack external validity.

Alternatively, time to stabilization (TTS) is a postural control measure that is used in conjunction with a functional jumping protocol. Time to stabilization is a measure of neuromuscular control and incorporates feedback from sensory and neuromotor systems during jump landings as the body transitions from a dynamic to static state (Wickstrom et al., 2005). Time to stabilization is also more reliable than other options such as center of pressure measurements (Wickstrom et al., 2006). The purpose of the present study was to examine the acute effect of WBV on bilateral and unilateral balance during jump landings of women basketball players, using TTS.

METHODS: Eleven current NCAA Division I women basketball players (age: 19.82 ± 2.18 years; height: 178.36 ± 10.64 cm; mass: 73.77 ± 15.29 kg) volunteered to serve as subjects. The study was conducted during the sport season and all subjects were participating in basketball practice, games, and strength and conditioning programs. Subjects gave informed consent before participating and human subject's research approval was obtained from the university office of research compliance before beginning the study.

Prior to the habituation and test sessions, subjects performed a warm-up, dynamic stretching, and vertical jumps of increasing intensity. The habituation session consisted of assessing CMJ height in order to provide subjects with an overhead target to use during the test sessions. Overhead targets were used since they have been shown to maximize CMJ height (Ford, et al., 2005). Subjects were taught and practiced the bilateral, right leg, and left leg CMJ and landings. This process included jumping to a maximum height, landing in an athletic position characterize by the shoulders over the knees, bringing arms down to a ninety-degree angle and "locking in" place upon landing, stabilizing as quickly as possible, facing straight ahead, and remaining motionless for a period of 5-7 seconds, consistent with methods previously used (Flanagan et al., 2008). Subjects also practiced the vibration protocol to be used during the test.

The order of the non-vibration and vibration test conditions was counterbalanced with each condition separated by 48 hours. Subjects then performed two trials of bilateral, right leg, and left leg CMJ, with the average of the two trials used for analysis. A one minute rest interval and randomization was provided between the bilateral, right let, and left leg CMJ to reduce order and fatigue effects.

A WBV platform (PowerPlate® Model PP040556129, PRO5 AIRDAPTIVE™ High Performance, Irving, CA, USA) was used as a vibration-loading device. The duration of the vibration stimulus was two minutes. During the first 30 seconds, subjects stood in an athletic position with slight knee joint flexion, shoulders over the knees, feet shoulder width apart, and a vibration frequency of 30 Hz was used. During the next 60 seconds subjects performed 10-12 slow continuous bodyweight squats at a vibration frequency of 40 Hz. For the final 30 second interval, subjects again stood in an athletic position, similar to that which was used during the first 30 seconds, but at a vibration frequency of 50 Hz. Vibration amplitude was 4 mm throughout.

After a minute, subjects performed two trials of bilateral, right leg, and left leg CMJ. Each CMJ was performed by taking off from and landing on a 60 x 120 cm force platform (BP60011200, Advanced Mechanical Technologies, Inc., Watertown, MA, USA). Kinetic data were collected at 1000 Hz, real-time displayed, and saved with the use of computer software (BioAnalysis 3.0, Advanced Mechanical Technologies, Inc., Watertown, MA, USA) for later analysis. Vertical ground reaction force data were collected for the sample period and used to calculate several variables from the vertical force components. Instants of initial foot contact, take-off, and landing were identified from the vertical ground-reaction force datasets. Vertical TTS was established as the time from the point of landing to when the vertical force component reached and stayed within 5% of the subject's body weight for one second.

All statistical analyses of the data were carried out in SPSS 16.0. A one-way ANOVA was conducted to assess differences in TTS between the vibration and non-vibration condition for the bilateral, right leg, and left leg CMJ. All data are expressed as means ± SD, with statistical effect size (d) and power (η_p^2) are reported. The *a priori* alpha level was set at $p \leq 0.05$.

RESULTS: Results of the statistical analysis revealed no significant difference in TTS between the vibration and non-vibration conditions for the bilateral ($p = 0.24$, $d = 0.20$, $\eta_p^2 =$

0.13) and right leg (p = 0.48, d = 0.10, η_p^2 = 0.05) CMJ. A significant difference in TTS was found between the vibration and non-vibration conditions for the left leg CMJ (p = 0.039, d = 0.57, η_p^2 = 0.36). Table 1 demonstrates the TTS data for vibration and non-vibration conditions.

Table 1. Time to stabilization (TTS) expressed as mean ± SD seconds (seconds) for the vibration and non-vibration conditions of each of the countermovement jump (CMJ) variations, and the difference between conditions.

	Vibration	Non-Vibration	Difference
Bilateral CMJ TTS (sec.)	1.00 ± 1.02	0.65 ± 0.15	35.0%
Right leg CMJ TTS (sec.)	0.90 ± 0.36	0.80 ± 0.30	11.2%
Left leg CMJ TTS (sec.)	1.25 ± 0.65	0.81 ± 0.29	35.2%*

*Significant difference ($p \leq 0.05$)

DISCUSSION: This study demonstrates that acute WBV, as performed in this study, offers no advantage and may impair dynamic stability. The TTS in this study was similar in some cases to the values of 0.97 seconds found by Flanagan et al. (2008). The TTS values found in the present study where shorter than the TTS values of 2.2 seconds demonstrated by Wickstrom et al. (2004) who studied a horizontal jump landing, compared to the vertical landings used in this study. In the present study, large TTS standard deviations in the vibration condition of the bilateral CMJ suggest that subject response to the vibration stimuli is highly variable, especially when compared the relatively small standard deviations found in the non-vibration condition.

Previous research has evaluated the nature of the vibration stimulus demonstrating potential optimal combinations of frequency and amplitude and it effects jump performance and power output. This research indicates that low frequency and amplitude or high frequency and amplitude combinations may be most effective (Adams et al., 2009). The present study used a range of low to high frequency stimuli and moderate displacement of 4 mm. Thus, it is possible that the vibration stimulus used in the present study may not have been optimal.

Vibration may enhance neuromuscular performance in athletes by increasing the muscle activity via muscle spindles (Cardinale & Bosco, 2003). Muscle spindles are the basis of the myotatic reflex, which is an important part of regulating posture. The myotatic reflex keeps length of a muscle constant (Robergs & Roberts, 1997). If muscles spindles are over-stimulated, the result is fatigue (Martin & Park, 1997). Torvinen et al. (2002a) showed that with a 4 min WBV session the surface electromyography (EMG) decreased in the vastus lateralis and gluteus medias during the vibration, which may indicate fatigue in those muscles. While some evidence indicates that the duration of the WBV session does not matter (Adams et al., 2009), it is also possible that the duration of the WBV stimulus in the present study may have been too long.

Time to stability provides researchers with a mechanism with which to assess dynamic stability during jump landings for basketball players, who demonstrate decreased proprioception, stability, and reaction time, particularly when recovery from ankle injuries (Fu & Hui-Chan, 2005). Overall, the above findings of this study indicate that some prescriptions of acute WBV training may impair dynamic stability.

CONCLUSION: Results of this study demonstrate that in some cases, WBV does not enhance and may acutely impair dynamic stability and balance in collegiate Division I female basketball players. Other combinations of frequency, amplitude, and duration of the vibration stimulus may be more effective.

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THE EFFECT OF SQUAT DEPTH ON MUSCLE ACTIVATION IN MALE AND FEMALE CROSS-COUNTRY RUNNERS

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KEYWORDS: EMG, resistance training, strength

INTRODUCTION: The squat is a closed-chain lower body exercise that is regularly performed by many athletes. The squat has been shown to increase strength of the rectus femoris, biceps femoris, gastrocnemius (Isear et al., 1997) and erector spinae (Nuzzo et al., 2008). Squats of different depths have been shown to alter muscle activation in male weight lifters (Caterisano et al., 2002), but the findings may not be directly applicable to runners. Therefore, we chose to examine both male and female runners and and multiarticular muscles that often fatigue while running. Muscle activation during parallel and partial squats has not been examined in runners. Hanon et al. (2002) reported that the rectus femoris and biceps femoris are among the first muscles to fatigue in runners. The gastrocnemius becomes increasingly important for running uphill (Sloniger et al., 1997), and the lumbar erector spinae can help runners to maintain upright posture and decrease the risk of injury to the hamstrings (Hoskins & Pollard, 2005).
The purpose of this study was to determine the effect of squat depth on muscle activation in both male and female collegiate cross-country runners. This may help athletes and coaches to determine which squat depth is most effective. We hypothesized that the parallel squat would increase extensor muscle activity (i.e. hamstrings and erector spinae). Furthermore, we sought to determine if changes in muscle activity were different between males and females.

METHODS: Twenty Division I cross-country runners, 10 males (mean ± SD; age = 19.2 ± 1.2 years, height = 176.8 ± 4.8 cm; body mass = 66.2 ± 8.0 kg; bodyfat percentage = 9.0 ± 3.5 %) and 10 females (age = 19.9 ± 1.2 years, height = 166.7 ± 4.7 cm; body mass = 55.9 ± 4.4 kg; bodyfat percentage = 19.7 ± 4.2 %) volunteered to serve as participants for the study. Informed consent and Institutional Review Board approval were obtained prior to the study.
Participants completed an orientation session that included body composition assessment, detailed instructions for the partial and parallel squats, joint angle assessment via a standard goniometer, and a 10 repetition maximum (RM) assessment for each squat condition. Before testing for their 10 RM, participants performed a warm-up of light cycling for 5 minutes on a stationary bike followed by 2 minutes of rest. Each participant's 10RM was then determined within 3 sets.
Electromyography (EMG) testing occurred within 7 to 10 days after the orientation session. The order of trials on the EMG testing day was randomized by a coin toss with heads indicating partial squat first and tails indicating parallel squat first. Participants completed 6 repetitions for each squat condition in a randomized order with their 10 RM loads. Partial and parallel squats were designated as 45 and 90˚ at the knee joint respectively. Repetitions were paced by a metronome set at 60 Hz. Cadence was 1 second down and 1 second up for the partial squat and 2 seconds down and 2 seconds up for the parallel squat. Electromyography was performed on the right rectus femoris, biceps femoris, lumbar erector spinae, and lateral head of the gastrocnemius during each condition. A BioPac Systems (Goleta,CA) EMG unit was used to record muscle activity. Sampling rate was 2000 Hz, and all data were integrated using the root mean square method and averaged over 100 samples. High-pass and low-pass filters were set at 30 and 500 Hz respectively.
Data were analyzed with SPSS 17.0 using repeated measures ANOVA procedures with gender and squat condition as fixed factors. Significant differences were set as $p < 0.05$.

Table 1. Subject Descriptive Characteristics.

Characteristic	Partial Squat	Parallel Squat
10 RM (kg)	78.4 ± 20.4	51.2 ± 14.0*
Hip Joint Angle (degrees)	50.0 ± 12.5	94.6 ± 16.2*
Ankle Joint Angle (degrees)	77.7 ± 7.9	69.7 ± 7.9*

*Significantly different than partial squat condition ($p < 0.01$). Joint angles were measured with a standard goniometer during the orientation session, and knee joint angles were confirmed during EMG testing.

RESULTS: Rectus femoris and erector spinae activity were significantly higher during the parallel squat condition ($p < 0.05$). Biceps femoris and gastrocnemius activation was similar between the partial and parallel squats. No significant differences existed between males and females when examining the interactions between squat condition and gender.

Table 2. Millivolts of Muscle Activity Determined by EMG.

Muscle	Partial Squat	Parallel Squat
Rectus femoris	0.142 ± 0.050	0.177 ± 0.065*
Biceps femoris	0.066 ± 0.044	0.075 ± 0.056
Erector Spinae	0.133 ± 0.050	0.163 ± 0.050*
Gastrocnemius	0.053 ± 0.018	0.049 ± 0.018

*Significantly different than partial squat condition ($p < 0.05$)

DISCUSSION: The primary findings of this investigation are that the rectus femoris and erector spinae activity are significantly higher during the parallel squat when compared to the partial squat. This increase in muscle activation can be attributed to greater ranges of motion at the hip, knee, and ankle joints. Because the rectus femoris fatigues early (Hanon et al., 2002) and the erector spinae aids in maintaining an upright posture (Hoskins & Pollard, 2005), these findings could be of importance to runners.

Despite using a significantly lighter load during parallel squats, rectus femoris and erector spinae activity increased. Parallel squats could benefit runners by reducing compressive forces on the spine while maintaining, or increasing, muscle activity compared to partial squats. Runners may avoid poor running technique and premature fatigue of the rectus femoris by performing parallel squats (Hanon et al., 2002). By increasing erector spinae strength during the parallel squat, runners can benefit by maintaining more upright postures. Weakness in erector spinae can contribute to excessive trunk flexion and exacerbate the risk of hamstring injury during the terminal swing phase of running (Hoskins & Pollard, 2005).

CONCLUSION: Cross-country runners should focus on performing parallel squats to maximally activate the rectus femoris and erector spinae muscles. This increase in muscle activation can be achieved while using a reduced load during the parallel squat versus the partial squat.

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SPEED, STRENGTH & POWER CHARACTERISITCS OF HORIZONTAL JUMPERS

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KEYWORDS: speed, strength, power, jumping.

INTRODUCTION: It is well established that approach speed on the long and triple jump runways is the single-most important determinant of performance across a wide range of ability levels (Hay, 1986; Hay, 1992). However, the relationship between speed and jump distance decreases when the range of performances is reduced.
At an elite level speed is regarded as a pre-requisite, the differentiating factor between performances relates more to how well athlete control their speed when they make contact with the take-off board (and subsequent take-offs in the triple jump). In the take-off athletes typically experience vertical impact forces in the range of 7.9 to 12.6 x BW (Ramey and Williams, 1985), with ground contact times ranging from 120 to 180 ms (the higher values relating to the step and jump take-offs). It is therefore imperative that horizontal jumpers are conditioned appropriately to accept such high loading forces, be powerful and reactive, in order to generate vertical speed in such a small timescale. Graham-Smith and Lees (2002) suggested that performance is made up of three main interacting factors; speed, strength and technique (with power being a derivative of speed and strength). They added that optimal performance can only be achieved when all three factors are in ‘balance’.
The aim of this study was to develop a battery of tests to monitor speed, strength and power for horizontal jumpers in as functional a way as possible. The relationship between strength and power variables with speed and controlled functional performance was also investigated.

METHOD: Seventy five athletes (37 male, 38 female) from the UKA long and triple jump squads underwent a battery of tests to measure speed, strength, power and functional performance. Speed was assessed using a Laveg LDM 300C speed gun (Jenoptik) sampling at 100 Hz, recording 0-20m, 0-40m and 20-40m split times. Strength and rate of force development (RFD) was assessed via an isometric squat on a Kistler force platform (type 9286AA) sampling at 1000Hz for 5 seconds and adopting a position that reflected the body position in mid take-off (upright trunk, knee angle at 135 degrees and bar, hip and ankle being in vertical alignment). The peak force and the average RFD over 150ms were recorded. Power was assessed in terms of jump height when performing squat, countermovement (CMJ) and drop jumps (DJ) from heights of 20cm and 40cm on the force platform (sampling at 1000 Hz. Jump height was calculated on the basis of flight time using the formula, height jumped = $gT^2/8$ (where T = flight time). Reactivity Index in the drop jumps was calculated by the flight time divided by contact time. Contact was defined as the time when the vertical force exceeded 20N. Functional performance was assessed in controlled field based tests including a standing long jump and a 4 bounds + jump into the sand pit. The best of 2 attempts were recorded. All athletes included in the study had at least one familiarisation session, data presented here is for the months of November and December collected over a 5 year period from 2004 to 2009.

RESULTS: Normative data for male and female jumpers are presented in table 1. When both male and female data were combined the strength and power variables that were most strongly associated with sprint time (0-40m) were peak isometric force (R^2 = 0.65), RFD (R^2 = 0.33), CMJ height (R^2 = 0.64) and DJ40cm height (R^2 = 0.43). The same variables were also strongly associated with performance in the standing long jump; peak isometric force (R^2 = 0.64), RFD (R^2 = 0.48), CMJ height (R^2 = 0.76) and DJ40cm height (R^2 = 0.61), and the 4 bounds + jump test; peak isometric force (R^2 = 0.66), RFD (R^2 = 0.49), CMJ height (R^2 = 0.69) and DJ40cm height (R^2 = 0.56).

Table 1. Normative data for male and female horizontal jumpers

		Male		Female	
		Mean	SD	Mean	SD
Body Weight	(N)	766	77	612	57
Speed 0-40m	(s)	4.82	0.12	5.45	0.16
Strength (Isometric Squat)					
Peak Force	(N)	4476	803	2727	588
Peak Force / BW		5.85	0.81	4.47	0.93
RFD (150ms)	(N/s)	8852	3057	5264	2695
Power (Vertical Jump Tests)					
Squat Jump	(m)	0.44	0.10	0.36	0.04
CMJ	(m)	0.52	0.07	0.39	0.05
Drop Jump 20m	(m)	0.48	0.08	0.37	0.05
RI	RI	2.77	0.68	2.47	0.54
Contact time	(s)	0.24	0.07	0.23	0.06
Drop Jump 40 cm	(m)	0.50	0.09	0.38	0.05
RI	RI	2.71	0.69	2.56	0.58
Contact time	(s)	0.25	0.07	0.23	0.06
Horizontal Jumps					
Standing LJ	(m)	2.99	0.18	2.44	0.16
4 Bounds + Jump	(m)	16.05	1.00	12.88	0.62

DISCUSSION: The study has provided useful normative data on a group of elite horizontal jumpers and gives an insight into the physical requirements of athletes. No previous study has profiled speed, absolute strength, explosive power (in terms of jump height), RFD and reactivity index of horizontal jumpers, and attempted to associate these attributes to more functional (and technical) tests of performance, i.e. the standing long jump and the 4 bounds + jump. The tests quantified attributes across the force-velocity spectrum and were specific in terms of typical force-length relationships, the types of muscle loading induced and the stretch-shorten cycle. Tests that included a stretch-shorten cycle and those that mimicked the correct force-length characteristics in jumping demonstrated stronger relationships with sprint and functional performance than purely concentric tests, i.e. the squat jump. Future studies should also measure eccentric strength and examine its function in the compression phase of the take-off.

CONCLUSION: The study has successfully profiled physical attributes of male and female horizontal jumpers across the force-velocity spectrum and can be used to monitor performance throughout the training cycle.

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RELATIONSHIP BETWEEN LOWER EXTREMITY STIFFNESS AND ECCENTRIC LEG STRENGTH IN HORIZONTAL JUMPERS

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KEY WORDS: stiffness, eccentric strength, jumping.

INTRODUCTION: Eccentric strength in the lower extremity has been identified as a key performance component in the horizontal jumps (Graham-Smith & Lees, 2005). Whilst isokinetic dynamometry provides a safe and reliable method of testing maximal eccentric strength, it is often criticised as being non functional due to testing at constant angular velocity and being an open kinetic chain movement (Baltzopoulos & Brodie, 1989; Augustsson & Thomeé, 2000). Therefore, eccentric leg strength measured during isokinetic testing may have limited transfer to functional performance. Lower extremity stiffness, such as vertical stiffness and knee joint stiffness (Farley et al., 1998) and knee joint moment, can be calculated during functional movements utilising force platforms, motion analysis and inverse dynamics. The purpose of this study was to examine the relationship between isokinetic eccentric leg strength and measures of lower extremity stiffness and knee joint moment during a single leg hop for distance test.

METHOD: Ten horizontal jumpers (7 male, 3 female) participated in the study. Prior to testing each athlete performed a thorough warm up consisting of 5 minutes cycling on an ergometer and 15 minutes of stretching. Concentric and eccentric peak gravity corrected torque of the quadriceps and hamstrings muscle groups were tested at 60 deg/s on a KinCom isokinetic dynamometer (Chattanooga Group, Inc., Hixson, TN) adopting the overlay method. Subjects were familiarised with the protocol and conducted warm up trials prior to data collection.

The two functional measures of lower extremity stiffness were determined via a hop and jump test on an AMTI (600400) force platform, whilst being tracked via a 10 camera Qualisys Pro Reflex and a Panasonic (NV-GS320) camcorder. Force data was sampled at 1200Hz and Qualisys captured motion at 240Hz. Subjects were required to hop from a 40cm high box onto the force platform positioned 1.5m away, and to jump as far as possible forwards. The distance jumped was determined via video analysis (Quintic 9.03, version 17) as the horizontal distance from the toe at take-off to the heel in the subsequent landing. Prior to performing the tests, each subject had cluster sets of 4 retro reflective markers placed on the thighs, shank and pelvis. The foot segment was defined by 4 markers placed on the calcaneous, 1^{st}, 3^{rd} and 5^{th} metatarsals. Static markers placed on the medial and lateral malleoli, medial and lateral femoral condyles, greater trochanter, ASIS, PSIS and iliac crest were positioned to define the proximal and distal ends of each segment. Force and motion data were collected in QTrack (Qualisys AB) and exported as C3d files. Each athlete performed 3 trials on both legs.

Data was processed in Visual 3d (C-Motion, Inc., Rockville, MD, USA). A Butterworth 4^{th} order zero lag filter was used to smooth the data adopting cut-off frequencies of 25Hz for motion and 15Hz for force. Joint moments were calculated using inverse dynamics via the link-model-base option. Knee joint stiffness was determined by the change in joint moment divided by the change in knee angle between the instants of touch-down and maximum knee flexion (Farley et al., 1998). Vertical stiffness was determined by the peak active force divided by the range of vertical displacement in the compression phase. Displacement of the centre of mass was calculated via double integration of the vertical acceleration graph. Average values of the 3 trials were taken.

RESULTS: Means (±SD) and Pearson's correlation coefficients between eccentric strength and stiffness measures for dominant and non dominant limbs are presented in Table 1.

Vertical stiffness was found to exhibit significant relationships with eccentric peak torque in hamstrings ($p<0.01$) and quadriceps ($p<0.05$) for dominant and non dominant legs. Knee joint stiffness was significantly related to eccentric peak torque in both quadriceps and hamstrings for the dominant leg, but not in the non-dominant leg. There was a significant relationship between jump distance and eccentric peak torque of both the quadriceps and hamstrings for the dominant leg ($p<0.01$), and eccentric peak torque of the quadriceps in the non dominant leg ($p<0.05$).

DISCUSSION: Peak knee joint moments in the present study were slightly greater than the 250-300Nm range previously reported during the long jump take-off (Stefanyshyn & Nigg, 1998). However, these differences are likely to be due to the initial hop from a height of 40cm in the present study, placing a higher level of eccentric loading upon landing. In isolation peak knee joint moment was not significantly related to eccentric peak torque of the quadriceps or hamstrings. However, it is an integral variable in the measurement of knee joint stiffness and this was significantly related to eccentric strength of the quadriceps and hamstrings, but only for the dominant limb (both $p<0.01$). This highlights the importance of limiting the range knee flexion to enhance stiffness when jumping. Interestingly vertical stiffness yielded the strongest relationships to eccentric leg strength for both limbs. This may be due to the fact that vertical stiffness reflects the stiffness of the entire lower extremity, particularly around the knee and hip joints. It is recommended that future studies quantify stiffness around the ankle, knee and hip joints, and examine their individual and combined relationships with eccentric strength and performance.

Table 1. Relationships between eccentric leg strength and lower extremity stiffness

	Dominant				Non Dominant			
			Correlation (r)				Correlation (r)	
	Mean	± SD	Ecc Quads	Ecc Hams	Mean	± SD	Ecc Quads	Ecc Hams
Pk Torque Ecc Quads (Nm)	331	71	-	-	325	66	-	-
Pk Torque Ecc Hams (Nm)	163	37	-	-	171	51	-	-
Knee Moment (Nm)	408	162	0.473 NS	0.434 NS	368	124	0.466 NS	0.541 NS
Knee Stiffness (Nm/deg)	13.1	6.7	0.871**	0.736*	9.7	5.2	0.411 NS	0.451 NS
Vertical Stiffness (kN/m)	16.7	5.3	0.815**	0.855**	15.0	5.6	0.671*	0.900**
Jump Distance (m)	2.06	0.33	0.759**	0.832**	2.12	0.33	0.688*	0.493 NS

** $p<0.01$; * $p<0.05$; NS Non significant

CONCLUSION: Vertical stiffness measured in a hop test from a drop height of 40cm produced the highest association with eccentric strength of hamstrings and quadriceps. From an applied perspective, monitoring of vertical stiffness using a force platform could provide an indication of eccentric strength of the lower extremity relatively quickly and in a more functional way than isokinetic testing.

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CORRELATION BETWEEN CLINICAL AND LABORATORIAL MEASUREMENT OF HAMSTRING FLEXIBILITY

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Maximal joint range of motion (ROM) represents the muscular flexibility level. However, different methods are used to measure the ROM at clinic and laboratory. So, the aim of this study was correlate a clinical and a laboratorial measurement of hamstring flexibility. The flexibility of both lower limbs of thirty-six young and healthy subjects was assessed by two apparatus: modified knee extension test (clinical measure) and Flexmachine (laboratorial measure). The results showed a moderate positive and significant correlation ($r=0.693$; $p<0.001$) and a common variance of 48% between the maximal ROM measured by these tests. It suggests that clinical and laboratorial tests are independent in nature. Further studies are necessary to correlate the maximal ROM clinically measured and others physiological and biomechanical variables.

KEYWORDS: Joint range of motion, flexibility measurement, electromyography, correlation, hamstring, stretching.

INTRODUCTION: Muscle flexibility is generally measured by the joint range of motion (ROM) and the maximal value of ROM represents the individual flexibility level (McHugh et al., 1998). Several clinical and laboratorial tests are used to measure the ROM. Quantitative criteria, such as onset of EMG activity are commonly used at laboratorial tests (Magalhães et al., 2007). This procedure enable estimate maximal passive ROM of the muscle tendon unit (MTU). However, the use of electromyography is not a common tool at clinical practice, what difficult the measurement of maximal passive ROM without the influence of muscle contraction.

Many authors perform clinical tests to estimate hamstring flexibility by knee extension ROM (Baltaci et al., 2003; Chagas et al., 2008; Roberts & Wilson, 1999). At these tests, ROM is usually registered by a goniometer and a subjective criterion is used to indicate maximal ROM, for example, highest passive resistance perceived by an examiner. Correlational research would provide insights into the relative importance of different tests for clinical practice and training control. We hypothesized that maximal ROM measures obtained at clinical and laboratorial test are independent. The purpose of this study was to correlate a clinical and a laboratorial measurement of knee extension ROM, which represents hamstring flexibility.

METHOD: Eighteen male and eighteen female volunteer to this study (mean ± SD; age: 24.2 ± 3.2 years; mass: 67.2 ± 13.0 Kg; height: 169.8 ± 7.9 cm), all free of any pathology in lower limbs, lower back or pelvis. Knee extension ROM was measured by two tests: Modified Knee Extension Test (MKET) (Chagas et al., 2008) and Flexmachine (Magalhães et al., 2007). Three measures were obtained at each test and the mean values were considered for both lower limbs. The intraclass correlation coefficient for MKET was 0,94 (Bergamini et al., 2005) and 0.92 for Flexmachine (Peixoto et al., 2007).

On the MKET, the subjects were positioned laid in supine on an adapted litter with a central cylinder and fixers on knee and hip to avoid compensatory movements. The volunteer's knee and hip were flexed at 90°, which was considered as 0° of knee extension ROM (figure 1). Next, the examiner extended the knee until achieve the highest passive resistance, which was determined by subjective perception. The ROM were recorded in this position by a flexometer (Leighton, Lafayette Instrument), and was defined as the maximal ROM (ROM_MKET).

Figure 1. Modified Knee Extension Test.

On the Flexmachine, an isokinetic machine composed by two chairs connected to a mechanical arm moved by a motor (SEW Eurodrive, Brazil), subjects were positioned at seated position with the hip flexed 45° from horizontal plane (figure 2). The center of subject´s knee was aligned with the axis of rotation of mechanical arm, which angle was continuously measured by a potentiometer fixed on this axis. Once triggered, the mechanical arm extended the knee joint passively (5º/s) until maximal stretch tolerance was achieved. For the measure, the 0° of knee extension ROM initiate at the vertical line from the ground. Electrical muscle activity was detected by Ag/AgCl surface electrodes (Midi-Trace® 2000 Foam, Canada) placed between the semitendineous and semimebranous muscles. The passive ROM (ROM_EMG) was obtained on the onset of the electromyogram signal (i.e. a raise bigger than two standard deviations of baseline mean) of the hamstring muscles during the test. The potentiometer values were acquired using analogical/digital convertor (Data Translation, DT BNC Box USB 9800 Series) and converted to angle by the software DASYLab 9.0 (Dasytech Laboratories, 32 bits). The Pearson product moment correlation coefficient was used to correlate the variables ROM_MKET and ROM_EMG and significance level of 95% was adopted.

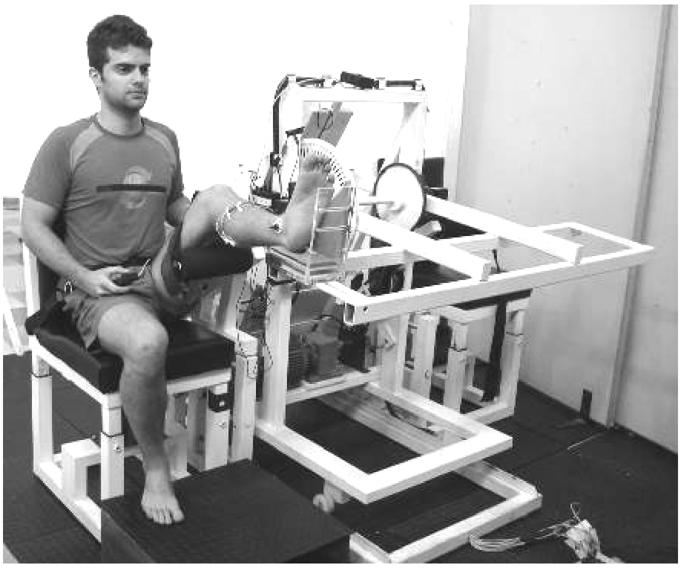

Figure 2. Flexmachine.

RESULTS: The descriptive data of ROM_MKET and ROM_EMG are present in table 1.

Table 1. Descriptive data of ROM_MKET and ROM_EMG.

	Mean	Standard deviation	Minimum	Maximum
ROM_MKET (º)	68.9	15.7	36	91
ROM_EMG (º)	93.2	20.2	51	132

There were a moderate positive and statistically significant correlation between the ROM_MKET and ROM_EMG (n=72; r=0.693; p<0.001). The coefficient of determination (R^2 x 100) was 48%. The results are presented in the figure 3.

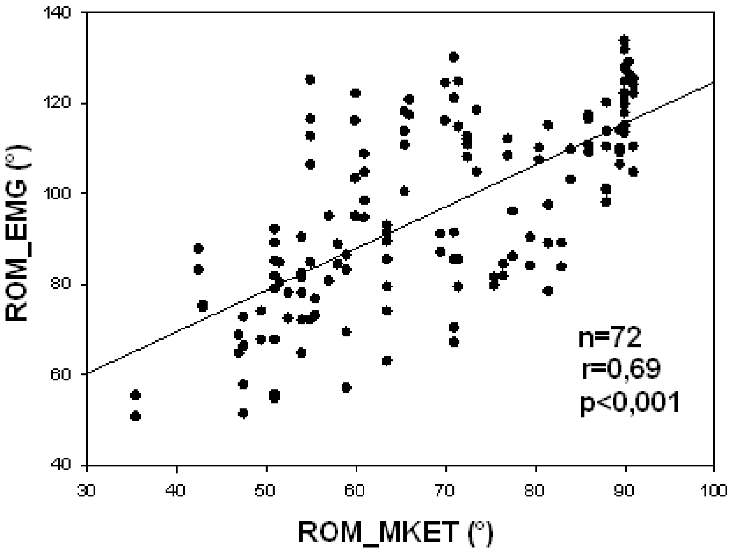

Figure 3. Scatter plot of ROM_MKET and ROM_EMG

DISCUSSION: The results demonstrate that knee extension ROM measured by MKET has moderate correlation with those obtained by Flexmachine. The coefficient of determination indicates that 48% of the variance in ROM_MEKT can be accounted for by knowing the variance in ROM_EMG. The others 52% reflect the proportion of variance that is not explained by the relationship between ROM_MKET and ROM_EMG. This outcome shows the use of ROM_MEKT as a possible predictor will result in a reasonable estimate of ROM_EMG, but not thoroughly accurate. This finding confirms previous report that found low relationship between clinical and laboratorial measures.

Blackburn *et al.* (2004) encountered a low correlation (r=0.36) between maximal ROM and passive stiffness. In this study, the maximal ROM was measured by very similar clinical test used in the present study and the passive stiffness was calculated using data obtained in laboratorial test. This indicate that the isolated measurement of maximal ROM in clinical tests not provide information about important characteristics of MTU during stretching. The shared common variance of only 48% shows that clinical and laboratorial tests are independent in nature. It suggests that differing physiological and biomechanical factors contribute to maximal ROM.

Further correlational approach could evaluate the relationship between other variables of flexibility, like passive torque, passive stiffness and work absorption, and maximal ROM obtained by clinical test. Other studies also are needed to evaluate the possible influence of the criteria used for determine maximal ROM, such as subjective criteria of the subject or of the examiner.

CONCLUSION: There is a moderate and significant correlation between the clinical (MKET) and the laboratorial (Flexmachine) measurement of knee extension ROM, however low common variance between both tests was found.

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ACUTE EFFECTS OF STRENGTH TRAINING ON RUNNING ECONOMY

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Sequencing strength training before aerobic conditioning is practised without empirical support. This study explored the acute effects of strength training on running economy and 3-D kinematics in five males. Running was performed on a treadmill at 12 and 14 km/h on three separate occasions. Trial 1 and 2 involved no strength training with data used to assess response stability of the variables. Before Trial 3, three sets of three repetitions at 85% of 1 repetition maximum of squat, bench press and deadlift with 3-5 minutes of rest were performed. Compared to Trial 2 no significant differences were observed when strength training was performed. Only a tendency of increased knee flexion (4.5°) at foot strike at the higher running velocity was observed. This suggests that running kinematics were changed exposing participants to long-term chronic injuries.

KEYWORDS: kinematics, fatigue

INTRODUCTION: When endurance training is performed first, studies have shown that post-activity strength and power measurements are significantly impaired compared to baseline data (e.g., Gomez et.al., 2002; Lepers, Pousson, Maffiuletti, Martin, & Van Hoecke, 2000). The time for full restoration of strength measures can take at least eight hours following the end of the endurance training (Gomez et al., 2002). Based on this prolonged time period required to recover, some authors recommend organising strength training before aerobic conditioning to avoid the aforementioned negative effects (e.g., Zatsiorsky & Kraemer, 2006). The practise of organising training in such a manner is now becoming popular within the professional team sport domain. The purported advantages are that it permits training to be completed under conditions more appropriate to improve strength in athletes and that both strength and endurance sessions are performed within a relatively short time-frame affording more time for the athletes to recover for the next day. Yet, within the literature there is a lack of supporting evidence for such practise. Moreover, when the strength training workload is too heavy (e.g., high volume typically used for hypertrophy) it can also illicit negative effects on proceeding actives. To date no previous research has investigated such decrements in running demands and movement coordination following a strength training protocol. From running induced fatigue it is known that movement coordination of running can change, increasing energetic demands and injury risk during the aerobic training (Derrick, Dereu & Mclean, 2002). This preliminary study explored the acute responses to a high-intensity, low volume strength training session in terms of running economy and kinematics. The aim was to identify any changes caused by the strength training to support or refute the emerging practise of organising strength training before endurance focussed conditioning.

METHOD: Five males (age = 23.6 ± 4.8 years; stature = 173.2 ± 5.8 cm; mass = 70.9 ± 1.3 kg; 1RM squat = 120 ± 9 kg; 1RM bench press = 80 ± 18 kg; 1RM deadlift = 112 ± 16 kg) with a minimum training age of six months volunteered. All were screened for injuries and medical conditions which could deem unfit for testing. For all trials they wore the same pair of running shoes and to standardise metabolic states during the testing they were asked to refrain from each of the following for the indicated time periods before testing: eating (≥2 hours), caffeine ingestion (≥4 hours), and vigorous or uncustomary exercise (≥24 hours) (Turner, Owings & Schwane, 2003). The testing schedule is shown in Figure 1. The maximum strength test included 3RM for the bench press and squat. These values were then used to estimate a 1RM for the bench press and squat (Wathen, as cited in LeSuer et.al., 1997, p. 211). Upon determination of the 1RM squat, another calculation was used (Ebben et.al., 2008) to determine the 6RM value for the deadlift. The 6RM values and Wathen's equation were again used to determine the 1RM value for the participant's deadlift. In total, three testing sessions were performed. Trial 1 and Trial 2 (completed with at least 48 hours between) were used for reliability purposes. Trial 3 consisted of a strength training program and occurred at least 48 hours after the second testing session. Immediately following the intervention session, participants performed another testing session on the treadmill

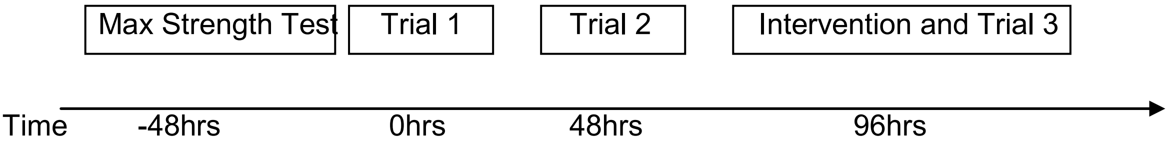

Figure 1. Time course of testing procedure.

Upon arriving for Trial 1, anthropometric characteristics were measured as described by Norton et.al. (1996, p. 33-75). This included stature, mass, and limb lengths. These measurements were later entered into the VICON Nexus (version 1.2) computer program. Bony landmarks were identified for the placement of retro-reflective ball markers for kinematic assessment using the Vicon MX motion analysis. Six Vicon cameras captured (500Hz) 3D motion of the participants when running on the treadmill. The study used the Plug-in-Gait lower limb marker set. A metabolic cart (MOXUS Modular VO2 System) was used to obtain running economy by dividing the VO_2 consumed during the final minute of each running stage by the participant's body mass. All running was performed on a Cosmos treadmill. Treadmill velocities were calibrated prior to commencement of testing, to ensure that the belt revolves at the same velocity as that displayed on the control panel. The participant then followed an 11 minute treadmill running protocol, with the treadmill set to a 1% grade incline to reflect the energy cost of outdoor running (Jones, 2007). The 11 minute protocol involved 3 mins at 10 km/h; 4 mins at 12 km/h; and 4 mins at 14 km/h. Data collection began in the final minute of each stage, where pulmonary gas exchange and lower limb kinematics were collected using the relevant equipment. For Trial 3, resistance exercise order was squat, bench press and deadlift. To replicate a strength training session to improve maximal strength participants performed 3 warm-up sets at 60, 70 and 80% 1RM, and 3 working sets of 3 repetitions at 85% 1RM in order to maximally stimulate motor units. Finally, rest intervals of 3-5 minutes duration were used (Kraemer et al., 2002).
Data from the right side were analysed only. The first two trials were used to assess test-retest stability of the dependent variables using typical error (TE), the 95% confidence intervals (CI) and the smallest worthwhile change (SWC) calculated by multiplying the standard deviation of the trials by a value of 0.2. T-tests, TE (95%CI) and Cohen's *d* were used to assess differences between Trial 2 and Trial 3. Angle-angle plots were visually inspected for each participant, at each velocity, across the three trials.

RESULTS: As shown in Table 1, TE ranged between 2.2 - 6.8° for kinematic measures while it was 1.7 - 2.3 $ml{\cdot}kg^{-1}{\cdot}min^{-1}$ for O_2 consumption. The SWC for all variables were less than their respective TE. Furthermore, based on the TEs and their respective 95%CI, test-retest stability was relatively poor for all variables. For all variables, no significant differences were found between Trial 2 and Trial 3. Only the knee angle at foot strike displayed a notable difference when the mean difference (Trial 3-2) was compared to respective TE and ES. On inspection of the angle-angle plots for each individual no systematic changes between Trials 2 and 3 were observed.

Table 1. Descriptive statistics across all three Trials.

			Trial 1 Mean(SD)	Trial 2 Mean(SD)	Trial 3 Mean(SD)	**Trial 1-2 diff**	**SWC**	**TE**	**(95%CI)**	**Trial 3-2 diff**	**Effect Size**	
Foot Strike	12km/h	Hip (°)	41.8(4.4)	37.3(4.5)	38.6(5.1)	4.5	0.9	2.9	(1.7-8.3)	1.3	0.3	
		Knee (°)	19.1(6.0)	19.6(7.5)	21.4(4.3)	-0.5	1.2	3.4	(2.1-9.8)	1.8	0.3	
		Ankle (°)	6.7(3.8)	6.8(3.8)	7.5(4.1)	-0.1	0.8	3.7	(2.2-10.8)	0.7	0.2	
	14km/h	Hip (°)	42.2(5.4)	40.4(6.1)	41.4(5.8)	1.9	1.1	2.4	(1.4-6.9)	1.0	0.2	
		Knee (°)	17.9(6.7)	20.6(7.9)	25.1(5.2)	-2.6	1.3	2.3	(1.4-6.6)	4.5	0.7	*
		Ankle (°)	4.7(3.3)	6.0(4.6)	7.9(3.3)	-1.3	0.7	3.6	(2.2-10.4)	1.9	0.5	
Toe Off	12km/h	Hip (°)	-5.5(4.9)	-8.4(2.8)	-6.8(2.6)	2.9	1.0	3.8	(2.3-11.0)	1.6	0.5	
		Knee (°)	15.6(4.7)	13.7(2.2)	16.0(4.1)	1.9	0.9	4.1	(2.4-11.7)	2.3	0.6	
		Ankle (°)	-20.5(2.3)	-17.3(5.8)	-18.2(4.0)	-3.2	0.5	4.5	(2.7-12.9)	-0.9	-0.2	
	14km/h	Hip (°)	-8.3(4.4)	-10.0(2.7)	-9.0(3.0)	1.7	0.9	4.3	(2.6-12.3)	1.0	0.3	
		Knee (°)	11.8(3.4)	14.7(0.9)	15.1(3.3)	-2.9	0.7	2.2	(1.3-6.2)	0.4	0.2	
		Ankle (°)	-21.5(4.3)	-15.7(5.8)	-18.5(1.8)	-5.8	0.9	6.8	(4.1-19.6)	-2.8	-0.7	
VO_2 ($ml \cdot kg^{-1} \cdot min^{-1}$)	12km/h		44.1(1.8)	42.1(4.5)	43.2(2.3)	2.0	0.4	2.3	(1.4-6.5)	1.1	0.4	
	14km/h		48.0(4.6)	49.3(5.0)	49.7(2.5)	-1.2	0.9	1.7	(1.0-4.9)	0.4	0.1	

Note. SD = standard deviation, VO2 = volume of oxygen, SWC = smallest worthwhile change, TE = typical error, CI = confidence interval.

DISCUSSION: Strength training had no significant effect to subsequent measures of running economy at either velocity. However, at 14 km/h there was a tendency by the group to foot strike with a more (4.5˚) flexed knee. This observation suggests participants were fatigued, since similar (4.4°) changes in knee flexion have been previously reported when running under a fatigued state (Derrick et al., 2002). Previously a greater knee flexion at foot strike has been identified as a potential injury risk, since it increases tibial shock (Lafortune, Hennig & Lake, 1996). Given this increase in knee flexion and the purported increase in risk injury it was interpreted that strength training prior to running may place athletes at a greater risk of injury. Explanations for a lack of change found (other than sample size) in running economy may be due to the assessment. This sub-maximal protocol was employed to assess the oxygen cost, as it has consistently been identified as a more sensitive assessment than aerobic power (i.e., VO2max). In addition, running economy was selected instead of a maximal aerobic assessment based on the difficulty for controlling motivation when asking participants to perform three maximal tests within a short time-frame. Perhaps a more ‘performance’ based measure such as a time-trial reflecting aerobic training would have been a more appropriate measure. It should be noted that an individual’s genetics and training history is a significant factor determining how prone to fatigue an athlete may be (Chiu et.al., 2003).

CONCLUSION: Possible evidence for increased injury risk was observed following a high intensity, low volume, full body strength training session. This was demonstrated by an increased knee flexion observed at foot strike (only at 14 km/h running velocity). It is likely that following the strength training participants were experiencing fatigue that changed the movement coordination. Despite the change in movement coordination running economy was preserved. These findings warrant a further more in-depth analysis using a larger, more experienced group of athletes.

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A COMPARISON OF LOWER BODY ANGLES BETWEEN FREE HIGH PULLS AND A FIXED HIGH PULL APPARATUS

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KEYWORDS: high pulls, angles, Cormax.

INTRODUCTION: The majority of strength and conditioning programs for athletes are based on the Olympic lifting exercises. Olympic lifts generate explosive power through the lower body (Armstrong, 1993). There appears to be a relationship between resistance training exercises and bar path kinematics (Souza, Schimada, & Koontz, 2002). The resistance training program at the university used in this study had a piece of equipment called the Cormax ® Smith Machine Plus. The Cormax® Smith Machine Plus utilizes a barbell that is set in tracks which does not allow any horizontal bar movement. It also has a piston system that allows the athlete to throw and release the barbell. The pistons support the barbell and allows it to slowly drop back to the starting postion. The researchers were interested if the technique using this piece of equipment would be similar to the technique that is used with free weight high pulls. Therefore, the purpose of this study was to examine the lower body joint kinematics between the two methods of completing a high pull.

METHOD: Six senior football players at a Midwestern university agreed to volunteer for the study. The participants (mean age 22.2 $\pm$.75 years, mean height 182.5 $\pm$5.1cm, mean weight 107.2 $\pm$20.7kg) had an average of five years of Olympic training experience. The participants were randomly assigned to complete four days of the high pull exercise. There were two or three days between all testing sessions. Two of the days the participants completed the high pulls using the Cormax ® Smith Machine Plus and two of the days they completed the high pulls as normal with a free barbell. The testing was alternated so that half of the subjects used the Cormax ® Smith Machine Plus the first day and half used the normal technique the first day. The second day the participants used the other technique, the third day they used the same technique as the first day, and the fourth day they used the same technique as the second day. The participants completed 3 sets of 5 reps at 75% of their 1RM. Two minutes of rest was given between sets.
Markers were placed on each participant's right side at each of the following locations: greater trochanter of the femur, the center of the knee joint, and the lateral mallelous of the ankle. The participants were videotaped from the right side during the high pulls. Dartfish® was used to measure the hip, knee, and ankle angles at the end of the first pull and the end of the second pull (See Figures 1 and 2). Three separate univariate ANOVAs were used to test for significant differences between the angles at the hip, knee, and ankle angles. The Bonferroni method was used to protect against alpha level inflation, resulting in p<.0083. Coefficients of variation (CV) were calculated for the joint angle measurements. Eta squared values were used to examine effect size.

Figure 1. Position where angles were measured for the end of the first pull.

Figure 2. Position where angles were measured for the end of the second pull.

RESULTS: Significant differences were found at the ankle angle during the first pull and at the hip, knee and ankle angles during the second pull (See Table 1). Eta squared values for the end of the first pull were 0.23 (ankle), 0.006 (knee), and 0.0007 (hip). Eta squared values for the end of the second pull were found to be 0.03 for all three of the joints.

Table 1. ANOVA Results of the Angle Measurements

Pull		Cormax Mean(SD)	CV	Platform Mean(SD)	CV	F	p
Pull 1							
	Ankle	103.7(5.3)	4.0%	98.6(3.1)	2.6%	157.77	.0001*
	Knee	141.4(6.2)	1.0%	143.3(5.8)	0.9%	3.32	.0694
	Hip	102.6(9.8)	2.0%	101.7(7.2)	1.7%	0.47	.4950
Pull 2							
	Ankle	129.4(5.7)	1.0%	131.1(5.5)	1.4%	12.32	.0005
	Knee	162.3(6.8)	1.0%	163.8(6.5)	0.8%	16.03	.0001*
	Hip	169.7(5.2)	0.8%	169.6(6.9)	0.8%	17.69	.0001*

*p=<0.0001

DISCUSSION: The results of this study indicate that there are some significant differences in the lower body kinematics between the Cormax ® Smith Machine Plus and the normal free weight technique when completing high pulls. The ankle angle was found to be greater for the Cormax ® Smith Machine Plus at the end of the first pull indicating that the participants may have shifted their body back to allow the bar to clear their knees. The hip, knee, and ankle angles were found to be smaller at the end of the second pull indicating that the subjects could not reach as much extension in the lower body using the Cormax ® Smith Machine Plus. The angles were only measured at the end of the first and second pull; there could be other similarities or differences in joint angles during the rest of the motion.

CONCLUSION: The results of this study showed that there were some significant differences in lower body kinematics between the Cormax ® Smith Machine Plus and the normal free weight technique when using high pulls. Due to its design, the Cormax ® Smith Machine Plus has been reported to have some power output advantages due to the ability to throw and release the barbell. However, if the design results in significant differences in lower body kinematics from the platform high pulls, any possible power advantages may not be worth the changes in kinematics. The changes in kinematics when using the Cormax ® Smith Machine Plus could transfer to the platform high pulls and affect the athletes' ability to properly perform the platform high pulls technique. This issue warrants further study.

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THE VALIDITY OF VELOCITY MEASUREMENT DURING UPPER-BODY RESISTANCE EXERCISES UNDER VARIABLE LOADS

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The purpose of this study is to compare the velocity of the barbell with the criterion of the velocity of the center of gravity during the explosive bench press lift with different loads. Fifteen highly trained soccer players participated in this study. Three-dimensional upper extremities kinematic data during the bench press were collected. Participants lifted loads of 10, 30, 50, 70 and 90% of one repetition maximum (1RM) with maximal velocity. All upper extremity segments and barbell were modeled as frustra of cones while the barbell was modeled as a cylinder. One way repeated ANOVA revealed that peak barbell velocity was significantly higher with 10, 30 and 50% and the mean barbell velocity was significantly higher with 10 and 30% of 1RM. Additionally, kinematic methods based on the barbell velocity measurement overestimate the velocity of the center of mass.

KEYWORDS: bench press, validity, center of gravity, velocity, resistance exercise, load power relationship.

INTRODUCTION: In the maximal effort strength training velocities used are one of the most important factors that determine the training stimuli and the consequent training adaptations. In addition, the velocity of the center of gravity influences the power output during the resistance exercise. Conditioners have used devices such as the linear position transducer or FitroDyne Premium (Slovakia) for measuring velocity during the strength training. The velocity of the barbell is supposed to be equal to the velocity of the center of gravity of the system lifted body mass and barbell. Thus, when using these methods for measuring velocity of the center of gravity, we hypothesize that the center of gravity of the system lifted body mass and barbell move in parallel with the barbell. This simplification may overestimate the velocity of the center of gravity and consequently the calculation of the power output. Moreover, the load-velocity and load-power relationship could be influenced as well. The previous study measured the barbell velocity instead of the velocity of the center of mass during the resistance exercise (Hori et al., 2007; Cormie, McBride, & McCaulley, 2007). The purpose of this study is to compare the velocity of the barbell with the criterion - velocity of the center of gravity (COG system upper-extremities and barbell) during the bench press lift with different loads. We hypothesize that the barbell velocity is significantly higher than COG velocity, especially with lighter loads.

METHOD: In the current study, fifteen professional male soccer players with a mean ± *SD* age, height, and body mass of 26.1 ± 3.9 years, 183.3 ± 6.7cm, and 78.8 ± 7.2kg, respectively, were tested for the maximum upper-body strength (one repetition maximum [1RM], bench press [BP]) and maximal velocity with various barbell loads. Three-dimensional upper extremities kinematic data during the bench press were collected at 247 Hz with a seven camera motion capture system (Qualisys Oqus, Sweden). The linear position transducer device (FitroDyne Premium, Slovakia) presented a sound that the subject could hear throughout the trial and that changed when the downward movement switched to the upward phase of the movement. The testing was performed with the use of free weight form techniques. Each subject visited the laboratory on two separate occasions with one-week rest. In the first session, the subjects were given instructions on the techniques of the bench press (Zatsiorsky, & Kraemer, 2006) and the range of motion for each subject was established without a chest-touch position. The body height, weight of the body and mass of upper extremity segments were then determined with the use of the segmental body composition analyzer (TANITA 418 MA, USA) (Kutač, & Gajda, 2009). The first session involved one repetition maximum testing according to the protocol published by Kraemer, Ratamess, Fry and French (2006). The second session involved measurement of the velocity while systematically increasing the load 10, 30, 50, 70 and 90% of one repetition maximum. Retro reflective markers were attached to the acromion, greater tubercle of humerus, medial epicondyle of humerus, lateral epicondyle of humerus, styloid process of radius, styloid process of ulna, terminal points of the barbell and the medial point of the barbell. Moreover, four light-weight rigid plates holding triads of markers were attached to the upper arms and forearms. After collecting a static trial in which they were required to stand in the initial upper position with the barbell, three acceptable trials in each load were collected. An acceptable trial was one in which the subject complied with the range of motion during the lift. The subjects were required to lift the load with maximal speed. A three-minute rest was given between each lift. Three trials in each load were collected. The trial with the highest mean velocity was accepted for further analysis. The marker data were processed using Visual 3D software (C-motion, Rockville, MD, USA). All upper extremity segments and barbell were modeled as frustra of cones while the barbell was modeled as a cylinder. The weight of upper extremities was calculated as a product of the mass of upper extremities and gravity acceleration. The velocity of the center of gravity (system upper-extremities and barbell) and the velocity of the barbell were the necessary parameters derived from the visual 3D software. For the statistical analysis, the signal of velocity was normalized to the time of lift. The peak velocity, mean, standard deviation (*SD*) and effect of the size were determined for the velocity for each load. One way repeated ANOVA was used to determine whether any significant differences in the vertical velocity existed between the methodologies. Additional comparisons were made between both techniques under different loading conditions in order to determine their effect on the load-velocity relationship. The reliability of the measurement was estimated by the intraclass correlation coefficient. The statistical significance was accepted at an α level of $p \leq 0.05$.

RESULTS: We found out that the mean barbell velocity was significantly higher than the mean COG velocity with 10 and 30% of one repetition maximum loads (Table 1). The peak barbell velocity was significantly higher than COG velocity with 10, 30 and 50% of one repetition maximum loads (Table 2).

Table 1. Comparison of the mean barbell velocity and COG velocity (system upper-extremities and barbell) for the bench press exercise (n=15)

Load (% of 1RM)	Velocity of Barbell Mean ± SD (m/s)		Velocity of COG Mean ± SD (m/s)		Δv (m/s)	ES
10∗	1.44	0.12	1.27	0.10	0.17	1.7
30∗	1.12	0.09	1.05	0.08	0.07	0.8
50	0.87	0.08	0.86	0.07	0.01	0.1
70	0.63	0.06	0.61	0.06	0.02	0.3
90	0.35	0.06	0.35	0.05	0	0

SD = standard deviation, Δv= difference in mean velocity, ES =effect of size

Table 2. Comparison of the peak barbell velocity and COG velocity (system upper-extremities and barbell) for the bench press exercise (n=15)

Load (% of 1RM)	Velocity of Barbell Peak ± SD (m/s)		Velocity of COG Peak ± SD (m/s)		Δv (m/s)	ES
10*	2.78	0.29	2.40	0.25	0.38	1
30*	1.96	0.12	1.81	0.12	0.15	1.25
50*	1.46	0.08	1.40	0.07	0.06	0.9
70	1.01	0.09	0.98	0.09	0.03	0.3
90	0.59	0.12	0.58	0.12	0.01	0.1

SD = standard deviation, Δv = difference in peak velocity, ES = effect of size

Figure 1 shows the effect of the methods and load used on the velocity curve during the normalized time of lift. The instantaneous barbell velocity optically has a similar behavior to COG velocity with all loads. The instantaneous barbell velocity optically shows higher values with all loads than the instantaneous COG velocity does. The curve of the instantaneous COG velocity with higher loads is closer to the curve of the instantaneous barbell velocity.

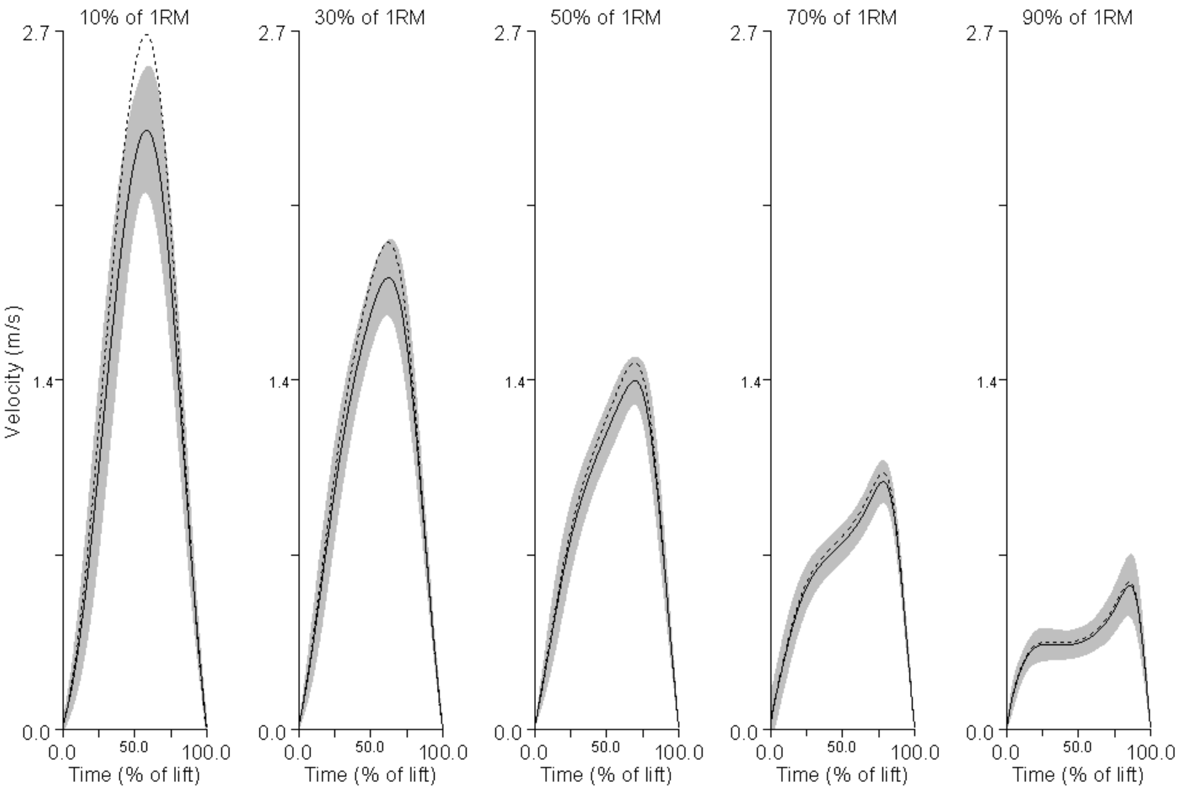

Figure 1. Normalized time-velocity relationship for different loads as derived by the two methods examined. The velocity of the barbell is expressed as a dot line and the criterion - velocity of the center of gravity (system upper-extremity and barbell) as a solid line. The gray area represents the between subjects standard deviation of the criterion (COG velocity).

Both methods displayed intraclass correlation coefficients (ICC) above the minimum acceptable criterion of 0.7 (mean barbell velocity between 0.715 and 0.929, mean COG velocity between 0.708 and 0.927, peak barbell velocity between 0.827 and 0.935 and peak COG velocity between 0.825 and 0.929) and were significant at an alpha level of 0.05 (Baumgartner, & Chung, 2001).

DISCUSSION: The purpose of this study was to verify the validity of the barbell velocity measurement during the bench press lift with different loads. We can confirm our hypothesis that the barbell velocity is not a precise estimate of the maximal velocity of the center of gravity during the bench press exercise. The barbell velocity is significantly higher with 10, 30 and 50% of one repetition maximum loads (Table 1 and 2). Based on these findings we could speculate that the load-power and load-velocity relationship is affected by this presumption of a parallel movement of the center of gravity and the barbell during the measurement of the velocity with a device such as FitroDyne Premium or Linear position transducer. Cormie at al. (2007) stated in their research that methods solely reliant on kinematic or kinetic data have limitations when used for the determination of the power output. Cormie at al. recommended a combination of the kinetic and kinematic (barbell velocity) measurement of the power output. This study demonstrated how the kinematic methods based on the barbell velocity measurement overestimate the velocity of the center of mass. These findings evoke a need to verify the effect of overestimating velocity on the load-power and load-velocity relationship.

CONCLUSION: The mean barbell velocity was significantly higher than the mean velocity of the center of mass (system barbell and upper extremities) with 10 and 30% (ES 1.7 and 0.8 respectively) of one repetition maximum loads during the explosive bench press exercise. The peak barbell velocity was significantly higher than the peak velocity of the center of mass (system barbell and upper extremities) with 10, 30 and 50% (ES 1, 1.25 and 0.9 respectively) of one repetition maximum loads during the explosive bench press exercise. Subsequent research could show the effect of overestimating velocity on the load-power and load-velocity relationship and consequently the optimal load to achieve the maximal power output during the bench press exercise.

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ANALYSIS OF THE TRAJECTORY OF CENTER OF MASS ON DIFFERENT SQUAT POSTURES AND LOADINGS

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KEY WORDS: weight training, free weight squats, knee joint.

INTRODUCTION: Weight lifting is usually used for lower extremity training, however, it might cause muscle or joint injury due to wrong posture. The center of mass (COM) could be measured as a parameter to monitor human movement. The purpose of the current study was to investigate the trajectory of COM at different postures and loadings of barbell squats weight lifting. We hypothesized that loading on the knee joint and trajectory changes of COM during 1/3 squat were smaller than 1/2 squat.

METHOD: Twelve females participated in this study. The mean age, weight, and height of the participants were 20.08±1.18 yrs, 57.92±4.87 kg, and 163.42±3.73 cm, respectively. A motion capture system with 10 cameras (MX 13^{+}, VICON, UK) was used to monitor and record the trajectories of the reflective markers on the special anatomical positions of the whole body at the sampling speed of 250 Hz. The subject was asked to perform 1/2 (knee flexion 90˚) and 1/3 (knee flexion 60˚) barbell squats with different loadings of 0%, 25%, 50% and 75% 1RM weighting in random order. The COM during squat was calculated by the movements of the segments defined by the reflective makers using Dempster's method. Two-way ANOVA with repeated measures and LSD post hoc were used for statistics. The significant level was set as $p < .05$.

RESULTS: The maximum knee joint flexion during 1/3 and 1/2 squats were around 80˚ and 100˚, separately (Table 1). The results showed that the flexion angle during 1/3 squat was smaller than that during 1/2 squat ($p<.05$). The trajectories and the displacements of COM during 1/2 and 1/3 squats at different loadings were showed on Figure 1 and Table 2, separately. The displacement of COM during 1/3 squat was smaller than that during 1/2 squat ($p<.05$).

Table 1. Maximum Knee joint angle (Mean±SD in degree) during 1/3 and 1/2 squats with different loadings.

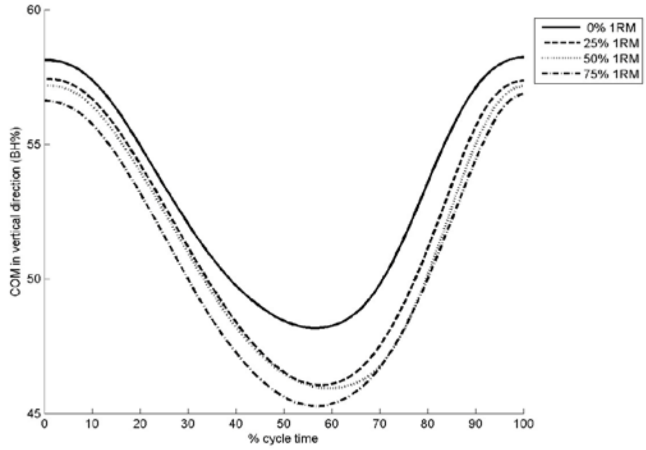
	0 %1RM	25%1RM	50%1RM	75%1RM
1/3 Squat*	73.81±11.33	80.20±11.45	79.99±11.09	81.14±13.78
1/2 Squat	100.06±11.87	100.41±12.97	99.89±10.50	96.35±12.15

* $p<.05$

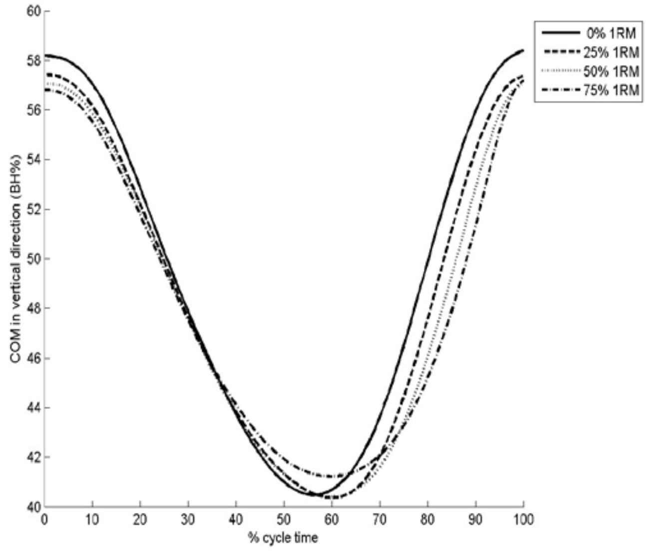
Table 2. Displacements (Mean±SD in %BH) of the COM during 1/3 and 1/2 squats with different loadings.

	0 %1RM	25%1RM	50%1RM	75%1RM
1/3 Squat*	10.20±1.44	11.60±1.83	11.50±1.23	11.60±1.65
1/2 Squat	19.39±6.91	17.37±2.49	16.99±2.57	16.05±2.40

* $p<.05$



(A)



(B)

Figure 1. The trajectories of COM during 1/3 (A) and 1/2 (B) squats with different loadings.

DISCUSSION: Lower loading on the knee joint and smaller displacement of the COM were noted during 1/3 squat in the current study. It indicated that the knee joint was less flexion and bore small moment and force. The more knee flexion angle, the greater knee joint loading (Scott, 1991). It might induce the knee injury if the joint was over loading. Therefore, free weight squats should be performed at less flexion angle to lower the joint loading of force and moment. To understand the muscle activity and joint power during squat, the joint moments and electromyography should be monitored and calculated in the future.

CONCLUSION: The results indicated that the maximum knee flexion angle of 1/3 and 1/2 squats were around 80˚ and 100˚, separately. The 1/3 free weight squat was suitable for lower extremity training because it had a knee flexion angle less than 90˚ and less displacement of the COM.

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EFFECT OF LOAD POSITIONING ON THE KINEMATICS AND KINETICS OF WEIGHTED JUMPS

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The present study sought to examine the effect of altering the position of external loads on the kinematics and kinetics of weighted vertical jumps in 29 resistance trained rugby union athletes. Vertical jumps were performed with loads of 20, 40 and 60% squat 1RM with the load positioned: 1) on the posterior aspect of the shoulder using a traditional barbell (TBJ); and 2) at arms' length using a hexagonal barbell (HBJ). Weighted jumps performed with the load held at arms' length resulted in significantly greater values for jump height, peak force, peak power, and peak rate of force development ($p<0.05$), indicating a greater training stimulus for the HBJ than TBJ. These results suggest that when using weighted vertical jumps to improve lower body muscular performance, the jumps should be performed with the external load at arms' length rather than on the shoulder.

KEYWORDS: Ballistic; power; weight-training.

INTRODUCTION: The vertical jump is an important feature of many sports and is often incorporated with other explosive body weight exercises in plyometric training programs aimed at developing muscular power and athletic performance. Resistance is frequently added to the vertical jump to increase the intensity of the training stimulus (Saez-Saez De Villarreal, 2009). The most common methods of applying resistance to the vertical jump include the use of dumbbells, barbells, weighted vests and elastic resistance bands secured to the floor. Research has shown that when resistance is applied to the vertical jump there is a significant change in the expression of force, velocity, power and rate of force production (Cormie, 2007; Moir 2005; Stone, 2003). Consistently, studies have demonstrated that the addition of resistance increases force production and decreases velocity and rate of force development (Cormie, 2007; Moir 2005; Stone, 2003). Varied results have been reported for the effect of added resistance on power production during vertical jumps. Some studies have reported that added resistance of approximately 40 to 60% 1RM may be required to maximise power (Stone, 2003; Sleivert, 2004). However, recent studies suggest that power is maximised when vertical jumps are performed unloaded (Cormie, 2007; Bevan 2010). If large external loads are included to alter the biomechanical stimulus of the vertical jump it is likely that positioning of the external load will be an important factor in determining the kinematic and kinetic changes. Difficulties exist in applying large resistances in the form of dumbbells, weighted vests or elastic bands. As a result, heavy loaded vertical jumps are customarily performed with the external load positioned on the posterior aspect of the shoulder using a straight barbell. An additional method which has not been considered in the literature is the use of the hexagonal barbell (Figure 1). The non-conventional barbell would enable the athlete to apply loads similar to that used during jumps with the weight placed on the shoulders. In addition, the design of the hexagonal barbell would enable the athlete to position the load closer to the body's centre of mass and reduce the resistance moment at the hip joint. As the weighted jump is considered one of the most effective methods to enhance lower body power (Baker 1996), the purpose of the study was to investigate whether changing position of the external resistance from the shoulders to arms' length would affect the kinematic and kinetics of the movement. Because this was an exploratory study, no formal hypotheses were specified.

METHODS: Twenty nine male rugby union athletes (age: 26.3 ± 4.6 yr; stature: 182.4 ± 6.8 cm; mass: 94.5 ± 13.1 kg; 1RM Squat: 153.7 ± 20.3 kg) gave informed consent to participate in this study, which was granted institutional ethical approval. All athletes had extensive resistance training experience and regularly performed weighted jumps in their strength & conditioning sessions. Data were collected for each subject over two sessions separated by one week. The first session involved 1RM testing in the squat and hexagonal barbell deadlift. The second session involved maximum effort jumps with loads of 0, 20, 40 and 60% of the recorded squat 1RM. Loaded jumps were performed across two conditions that altered the positioning of the external resistance. The first condition required the load to be placed across the posterior deltoids using a traditional barbell (TBJ). During the second condition subjects held the external resistance at arms' length using a hexagonal barbell (HBJ). All jumps were performed in a randomized order with two repetitions performed in each trial to assess reliability. Unloaded jumps (0% 1RM) were performed with the arms held stationary at the side of the body.

Trials were performed with a separate piezoelectric force platform (Kistler, Type 9281B Kistler Instruments, Winterthur, Switzerland) under each foot. Displacement, velocity and power data were calculated at the athletes' COM during unloaded trials, and at the system COM (athlete + external load) during loaded trials This was achieved by incorporating the vertical ground reaction force (VGRF) data and using the principle that the impulse applied to the system equals its change in momentum (Kawamori 2005). Briefly, trials were initiated with subjects standing erect and motionless. Once data acquisition was initiated, subjects were instructed to lower themselves to approximately 120° of hip flexion, where they then reversed the movement and attempted to jump as high as possible. Changes in vertical velocity of the system COM were calculated by multiplying the net VGRF (VGRF recorded at the force plate minus the weight of the athlete + external resistance) by the intersample time period divided by the mass of the system. Instantaneous velocity at the end of each sampling interval was determined by summing the previous changes in vertical velocity to the pre-interval absolute velocity, which was equal to zero at the start of the movement. The position change over each interval was calculated by taking the product of absolute velocity and the intersample time period. Vertical position of the system COM was then obtained by summing the position changes. The vertical velocity of the system at take-off was used to calculate jump height using the constant acceleration equation (Jump height = $(TOV)^2 / 2g$, where TOV = vertical velocity of the system COM at take-off, $g = 9.81 ms^{-2}$). Instantaneous power was calculated by taking the product of the VGRF and the concurrent vertical velocity of the system. Analyses were performed for the ascent phase only.

A general linear model with repeated measures and Bonferroni *post hoc* tests were used to determine significant differences. All statistical analyses were conducted using SPSS Version 15.0, with statistical significance accepted at a level of $p<0.05$

Figure 1: Weighted jumps performed with the traditional straight barbell and the hexagonal barbell.

RESULTS: Test-retest reliability for vertical jump height, average force, peak force, average velocity, peak velocity, average power, peak power and PRFD were all high (ICC = 0.98, 0.98, 0.97, 0.94, 0.90, 0.97, 0.94 and 0.80), respectively. Subjects were able to lift a significantly heavier 1RM load in the hexagonal barbell deadlift compared to the traditional barbell squat (195.4 ± 18.3 kg vs. 153.7 ± 20.3 kg, $p<0.05$). The mean jump height, peak force, peak velocity and peak power values for the group during the unloaded condition equalled 39.3 ± 5.5 cm, 1967 ± 202 N, 2.79 ± 0.18 ms^{-1}, 4324 ± 301 W, respectively. The addition of resistance to the vertical jump significantly increased peak force ($p<0.05$) and decreased peak velocity ($p<0.05$). Significantly greater peak power values were obtained for jumps performed with the hexagonal barbell and a load of 20% 1RM than all other conditions ($p<0.05$). Significantly higher weighted jumps were obtained when the external resistance was positioned at arms' length ($p<0.05$, Table 1).

Table 1. Loaded vertical jump heights obtained with 20, 40 and 60% 1RM barbell loads. *Significant difference of load position for corresponding load (p<0.05).

Condition	Vertical Jump Height 20% 1RM Mean ± SD	Vertical Jump Height 40% 1RM Mean ± SD	Vertical Jump Height 60% 1RM Mean ± SD
TBJ (cm)	20.2 ± 4.0*	14.0 ± 2.7	8.5 ± 2.1
HBJ (cm)	27.1 ± 3.9*	15.2 ± 2.6	8.9 ± 1.9

Significant main effects of load position were obtained for peak force, peak power, and peak rate of force development ($p<0.05$). For all variables measured there was a trend towards higher values when the external load was positioned at arms' length using the hexagonal barbell (Table 2).

Table 2. Kinematic and kinetic data for loaded jumps performed with 20, 40 and 60% 1RM barbell loads. *Significant difference of load position for corresponding load (p<0.05).

	20% 1RM Mean ± SD	40% 1RM Mean ± SD	60% 1RM Mean ± SD
TBJ Average Force (N)	1853 ± 214	2064 ± 204	2291 ± 201
HBJ Average Force (N)	1866 ± 164	2069 ± 171	2326 ± 163
TBJ Peak Force (N)	2243 ± 252*	2509 ± 233*	2726 ± 208*
HBJ Peak Force (N)	2353 ± 213*	2689 ± 252*	2945 ± 232*
TBJ Average Velocity (ms^{-1})	1.23 ± 0.12	0.98 ± 0.13	0.87 ± 0.15
HBJ Average Velocity (ms^{-1})	1.33 ± 0.11	1.03 ± 0.12	0.87 ± 0.07
TBJ Peak Velocity (ms^{-1})	2.28 ± 0.17*	1.94 ± 0.20	1.73 ± 0.21*
HBJ Peak Velocity (ms^{-1})	2.39 ± 0.18*	1.99 ± 0.16	1.76 ± 0.10*
TBJ Average Power (W)	1994 ± 224*	1857 ± 286	1590 ± 356
HBJ Average Power (W)	2158 ± 307*	2041 ± 279	1623 ± 202
TBJ Peak Power (W)	4091 ± 438*	4065 ± 508	3789 ± 542
HBJ Peak Power (W)	4606 ± 510*	4386 ± 544	3831 ± 345
TBJ PRFD (Ns^{-1})	4848 ± 1538*	4938 ± 924*	5085 ± 917*
HBJ PRFD (Ns^{-1})	27805 ± 2379*	9062 ± 2505*	8349 ± 2014*

DISCUSSION: Results demonstrated that positioning of the external resistance had a significant effect on the kinematics and kinetics of weighted jumps. Customarily, when athletes perform weighted jumps with substantial resistances the external load is positioned on the posterior aspect of the shoulder using a straight barbell. The current study shows that changing the position of the load from the shoulders to arms' length using a hexagonal barbell results in significant increases in peak force, peak power, and peak rate of force development, with a trend towards higher velocity, average force and average power values The results also demonstrate that the hexagonal barbell can be used to apply resistances equal to or greater than that obtainable with a straight barbell. As the analyses in this study were limited to the propulsive phase of the jump, future research should investigate whether positioning of the load effects landing kinematics and kinetics which are important factors for preventing injury.

Force and velocity data obtained in this study are similar to values reported previously for weighted jumps (Moir, 2005; Sleivert, 2004; Cormie 2007). Higher peak power values than those obtained here have been reported by Cormie et al (2007) and Sleivert et al (2003). Methodological differences in the calculation of power are likely to account for the variance in results. An extensive amount of research has been devoted to identifying loads that produce maximum power during vertical jumps due to the suggestion that these loads will be the most effective for developing power. When comparing unloaded and weighted jumps performed with the straight barbell in this study the results coincided with recent research (Cormie, 2007; Bevan 2010) showing that maximum power is produced during unloaded jumps. In contrast, when comparing jumps performed with the hexagonal barbell the results showed that maximum power was produced with a load of 20% 1RM, with no significant difference between the unloaded and 40% 1RM conditions. The results of this study should be used to inform the exercise and load selection for training sessions aimed at developing lower body power.

CONCLUSIONS: The weighted jump is considered to be one of the most effective exercises for developing lower body power. Traditionally, weighted jumps are performed with the external load positioned on the posterior aspect of the shoulder using a straight barbell. The results of this study show that a greater mechanical stimulus can be achieved by changing the position of the external load from the shoulders to arms' length through the use of a hexagonal barbell.

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Poster Session 2

ACUTE EFFECTS OF VERBAL FEEDBACK ON EXPLOSIVE UPPER-BODY PERFORMANCE IN ELITE ATHLETES

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The purpose of this study was to determine the effects of verbal feedback on the explosive upper-body performance of well-trained rugby union athletes in a resistance training session. Athletes (n = 9) completed two sessions of bench-throws with peak velocity feedback after each repetition, and two identical sessions without feedback. Within each session, three sets of four repetitions of bench-throw were completed. When feedback was received there was a small increase of 1.8% (90% confidence limits, ±2.7%) and 1.3% (±0.7%) in average peak power and velocity when averaged over the three sets. When individual sets were compared, there was a tendency towards greater increases in average peak power in the second and third sets. Benefits of feedback may be greatest in the latter sets of training and could improve training quality and adaptation.

KEYWORDS: Power, velocity, bench-throw, rugby union.

INTRODUCTION:

To be successful in a chosen sport, athletes need to develop a variety of specific skills and physical attributes. However in many sports, such as rugby union, athletes have limited time to train and develop each physical attribute before optimal recovery is compromised or injury risk is increased. It is possible the athletes sometimes train with insufficient motivation or intensity to maximise their training time. Therefore, improving the quality of each training session (without extending the duration or increasing the volume) is a common goal for many athletes, coaches and support staff.

Training quality can be affected by a number of factors, particularly motivation and intensity. Motivational strategies can be classified as either intrinsic (e.g. self talk or 'psyching up') or extrinsic (e.g. external feedback or encouragement) (Jung & Hallbeck, 2004; Tod et al., 2005). 'Psyching up' has been shown to increase isokinetic bench press strength by 11.8% when compared to a mental distraction control (Tod et al., 2005). Additionally, Jung and Hallbeck (2004) reported an increase in peak handgrip strength of approximately 5% when visual feedback or verbal encouragement were given. It should be noted that performance improvements in the aforementioned investigations were assessed in testing sessions consisting of a single repetition or set, an approach that is atypical in resistance training where multiple sets of multiple repetitions are performed consecutively (excluding one repetition maximum lifting). This is important as since regular resistance training sessions consist of multiple sets and repetitions, the fatigue which accumulates throughout the session will likely reduce the intensity of the work completed in the final sets, resulting in reduced training quality (Legaz-Arrese et al., 2007),. Therefore, the effect of motivational strategies on performance across a resistance training session still requires investigation.

While strength is important and often assessed in practice, research indicates that power may be a better predictor of athletic performance (Newton and Kraemer, 1994). Numerous authors have reported increases in lower body power when motivational strategies were implemented (Tod et al., 2009). To date, only one study has investigated the acute effects of motivation on upper body power (Antonis et al., 2004). These authors reported that intrinsic motivation (self talk) increased distance of an over head throw in water polo (7.2%) in untrained swim class students (Antonis et al., 2004). Thus, it is unknown whether verbal feedback can improve performance in upper body explosive exercises in well trained athletes. Therefore, the purpose of this investigation was to determine the effects of verbal feedback on upper body power in a resistance training session consisting of multiple sets and repetitions in well trained rugby athletes.

METHOD:
Nine elite rugby union athletes from a Super 14 professional rugby team volunteered to take part in this study during the competitive phase of their season (mean ± SD; age, 22.1 ± 2.1 years; height, 184.2 ± 7.7 cm; mass, 107.3 ± 13.2 kg; maximal bench press strength 135.9 ± 22.6 kg). Athletes were assessed using the bench-throw exercise on four separate occasions each separated by seven days. Each athlete completed two sessions consisting of three sets of four repetitions of the bench-throw with peak velocity feedback provided following each repetition; and two identical sessions where no-feedback was provided after each repetition. Each set was separated by two minutes rest. Athletes were randomly split into two groups which differed only in the order they received feedback or no-feedback over the four testing occasions (Figure 1).

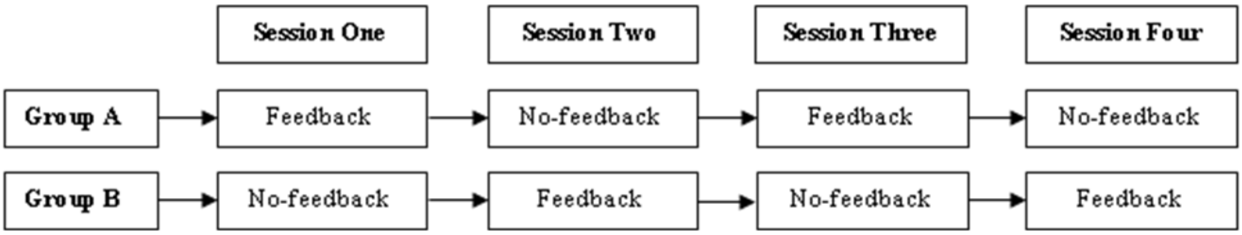

Figure 1. Outline of testing order. Group A, n=4; Group B, n=5.

A standardised warm up consisting of two sets of ten body-weight press ups followed by one set of five explosive press ups with a clap was completed. Athletes then completed three sets of four repetitions of bench-throws at a load of 40kg within a smith machine. Athletes used a self selected hand position and lowered the bar to a self selected depth (Argus et al., 2009). Athletes then threw the bar vertically and explosively as possible, trying to propel the bar for maximal velocity. In both conditions a one second pause occurred following the completion of each repetition (at the end of the concentric phase) so that peak velocity feedback or no-feedback could be provided (obtained via GymAware®; 50 Hz sample period with no data smoothing or filtering; Kinetic Performance Technology, Canberra, Australia). Athletes rested for two minutes between all warm up and training sets. Athletes were ask to rate their effort after each set; with all reporting maximal effort across all sets.
The first repetition from each set was excluded from analysis, as feedback could not be provided until after the completion of the first repetition. The repetitions for each set from the two feedback sessions were combined and averaged prior to analysis, as were the no-feedback repetitions. Average peak power and peak velocity data of all nine repetitions, as well as the average for each set of three repetitions (set one, two or three) were used for analysis.
All data were log-transformed to reduce non-uniformity of error, and the effects were derived by back transformation as percent changes (Hopkins et al., 2009). Standardised changes in the mean of each measure were used to assess magnitudes of effects by dividing the changes by the appropriate between-subject standard deviation. Standardised changes of <0.2, <0.6, <1.2, <2.0 and >2.0 were interpreted as trivial, small, moderate, large, and very large effects. To make inferences about the true (large-sample) value of an effect, the uncertainty in the effect was expressed as 90% confidence limits (Hopkins et al., 2009).

RESULTS:
A small increase of 1.8% (90% confidence limits; ±2.7%) in average peak power of all repetitions was observed when feedback was received. When each set was compared individually there was no difference in average peak power between the first set in either condition. The average peak power in the second set was 2.4% (±4.7%) greater when feedback was received compared to the second set of the no-feedback condition, representing a small effect. There was also a small increase of 3.1% (±3.3%) in average peak power of the third set in the feedback condition compared with no-feedback (Figure 2). Average peak velocity of all repetitions was 1.3% (±0.7%) greater when feedback was

provided and this represented a small effect. When each set was compared, a small improvement in average peak velocity was observed in all three sets in the feedback compared to no-feedback condition. Increases in average peak velocity were 1.3% (±1.1%), 1.1% (±1.1%) and 1.6% (±1.0%) for set one, two and three, respectively (Figure 2).

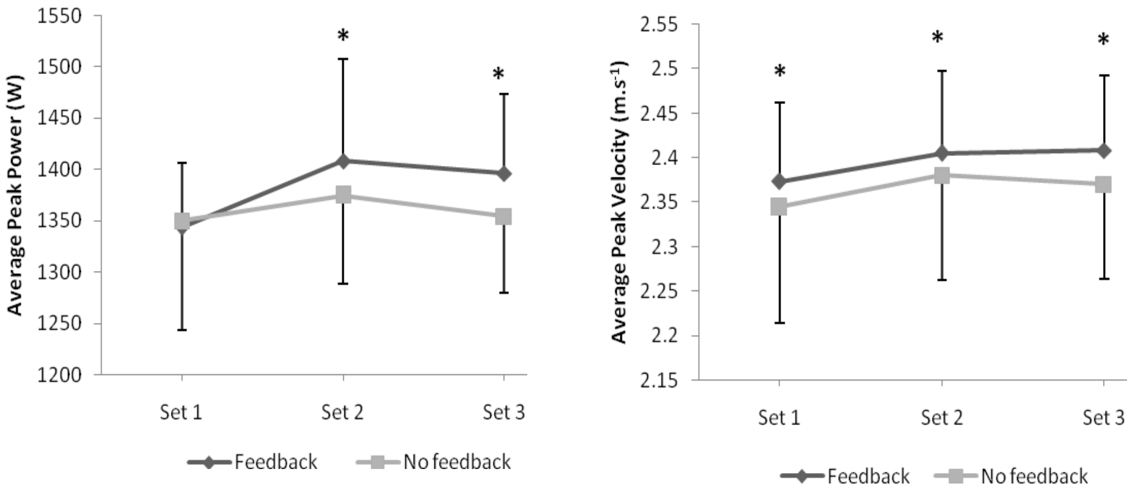

Figure 2. Average peak power and velocity with standard deviations (error bars) obtained during three sets of three repetitions of 40-kg bench throw. Peak velocity feedback was provided in a verbal manner at the completion of each repetition for the feedback condition. * denotes a small difference between conditions.

DISCUSSION:

Small improvements in bench-throw average peak power and average peak velocity were observed when specific feedback of performance was received immediately after each repetition. These results add to the current literature in several ways. Firstly, to our knowledge, only one other investigation has examined the effects of feedback on upper body power (Antonis et al., 2004). However, the previous investigation examined the effects of feedback on a complex technique based task (overhead throw), whereas the current investigation examined the effects of simple task (bench-throw). Secondly, this was the first investigation to examine the effects of feedback using assessment procedures typical of a traditional resistance training session i.e. consisting of multiple sets and repetitions. As such, the current investigation addresses a gap in the literature.

Receiving verbal feedback improved average peak power and velocity of the training session by 1.8% and 1.3%, respectively. The greatest benefit when receiving feedback appears to be in the latter sets of training. Indeed, when each set was analysed separately, improvements were greatest in the final set (3.1% peak power; 1.6% peak velocity). These findings suggest that receiving feedback improved the quality of training; which may be observed as an increase in, or maintenance of intensity or motivation in the latter sets of training. Additionally, if these improvements seen during one training session can be maintained over multiple training sessions, the long-term effects of repeating these "higher quality" sessions may result in enhanced training adaptations and therefore better performance (Kaneko *et al.*, 1983; Newton and Kraemer, 1994). Although the benefits gained may appear small, it should be noted that previous literature has reported 5% improvements in upper-body power in elite rugby league athletes over a four year period (Baker and Newton, 2006). As such acute improvements of ~3.1% in a single session are a positive and worthwhile finding.

Performance improvements were smaller than previously reported in studies investigating strategies that may alter motivation (Jung and Hallbeck, 2004; Tod *et al.*, 2005; Tod *et al.*, 2009). Differences may be due to the level of subjects and musculature recruited. It is commonly accepted that well trained individuals routinely evoke a greater percent of muscular activation than their untrained counterparts (Van Cutsem *et al.*, 1998). Therefore in untrained individuals, there may be greater potential for feedback to enhance muscular activation which may lead to greater performance improvements. Smaller improvements may

also be due to the muscle group involved (i.e. upper vs. lower body). The different response between the two extremities may be due to the disparity in the total muscle mass recruited, i.e. the larger muscle mass of the lower body may have greater scope for improvement.
The mechanisms for improvements as a result of feedback were not assessed in the current investigation. However, it can be speculated that improvements may be due to enhanced neuromuscular activation or increased intent. All athletes reported that maximal effort was given in each repetition on all testing occasions; nonetheless athletes still improved on their self reported maximal effort when feedback was received. As such, if athletes were indeed using maximal effort, it would suggets that feedback improved performance due to other mechanisms than motovation or increased intent. Further research should attempt to identify the mechanisms that lead to performance improvements with specific feedback.

CONCLUSION:
The use of verbal feedback resulted in acute increases in upper body average peak power and velocity. However it is unknown whether providing feedback to athletes will provide continuous acute adaptations in performance over a longer training phase, or if adaptation will diminish with repeated use. Providing verbal feedback increases the quality of training and produces acute improvements in power output of well trained athletes, in which small improvements are often difficult to achieve. Providing feedback may be particularly useful in periods where training volume is higher as the results of this investigation indicate that the benefit of feedback was greatest in the latter sets of training.

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COMPARISON OF JUMP HEIGHT VALUES DERIVED FROM A FORCE PLATFORM AND VERTEC

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This study simultaneously assessed jump heights derived from a force platform and a Vertec as well as the reliability of each instrument. Twenty-one recreationally active adults performed 3 maximal countermovement jumps reaching to a Vertec that was placed above the force platform. A repeated measures ANOVA was used to assess differences between Vertec jump height and force platform derived jump height. Results revealed a 27% higher jump height when assessed by the Vertec, compared to the force platform. Intra-class correlations were used to assess trial-to-trial reliability. Both instruments displayed excellent reliability. Practitioners could use the following regression equation to interpret measurements from the force platform: Vertec jump height = force platform height (1.024) + 0.142m.

KEYWORDS: countermovement jump, instrumentation, athlete testing, power, reliability

INTRODUCTION: The countermovement jump is performed in a variety of sports. Practitioners and sport scientists also use the countermovement jump to assess lower body power and the effectiveness of training protocols (Ham et al., 2007; Klavora, 2000).
A variety of instruments are used to evaluate countermovement jump performance including a measuring tape, Vertec, contact mat, motion analysis and force platform (Klavora, 2000; Leard et al., 2007). Jump height assessed by measuring tape and Vertec is determined by subtracting the standing height or reach height by the maximum jumping height or reach height using procedures such as Sargent's, Abalakov's, and Starosta's, jump tests (Klavora, 2000; Starosta & Radzinska, 2001). Jump height assessed by the force platform and contact mat can be determined by the time in air (Markovic et al., 2004; Moir, 2008). Jump height assessed by the aforementioned instruments have been shown to be reliable, however, significant jump height differences ranging from 7.9 to 36% have been reported (Isaacs, 1998; Leard et al., 2007; Markovic et al., 2004; Starosta & Radzinska, 2001). Practitioners must be able to interpret and compare results obtained by scientists that use laboratory equipment such as a force platform to a more inexpensive and convenient field test such as a Vertec.
Force platforms have been used to assess jump performance because of their precision based on high sampling frequencies, and accuracy when compared to motion analysis data (Baca, 1999; Mori, 2008). The Vertec measuring device is widely used because of its simplicity. This device requires the athlete to maximally reach for an object which closely replicates many common sport movements. This device is composed of 48 vanes spaced 1.27 cm apart which can be displaced by the hand when jumping and reaching to a maximum height. Jump heights assessed via Vertec have been shown to be 7.9 and 11% lower when compared to contact mats (Isaacs, 1998; Leard et al., 2007). However, previous research has yet to assess the differences in countermovement jump height obtained from a Vertec and a force platform. Thus, the purpose of this study is to assess the difference between Vertec and force platform derived jump heights, and to assess the reliability of each of the testing instruments.

METHODS: Twenty one recreationally active adults (six female and fifteen male; mean ± SD; age = 22.4 ± 5.1 years; height = 176.4 ± 9.1 cm; body mass = 78.6 ± 11.5 kg) volunteered to serve as subjects for the study. Inclusion criteria included subjects who were 18-45 years old, participated in high school or college sports, without orthopedic lower limb pathology that restricts functioning or known cardiovascular pathology. All subjects provided informed written consent and the study was approved by the institution review board.

Warm-up prior to the test consisted of three minutes of low intensity rope jumping followed by dynamic stretching including one exercise for each major muscle group. Three minutes rest was provided prior to beginning the test. The test consisted of three maximum countermovement jumps with arm swing, since this technique is sport specific and has been shown to maximize jump performance (Hara et al., 2008). Subjects were instructed to jump and reach maximally to a Vertec (Sports Imports, Columbus, OH, USA) which were performed on a force platform (BP6001200, AMTI, Watertown, MA, USA), thus enabling simultaneous recording of results by both methods. One minute rest was provided between each trial.

Jump height derived from the Vertec was assessed by the distance between the height of the highest vane touched during the standing vertical reach with one hand and the vane touched at the highest point of the jump with one hand measured to the nearest 1.27 cm (Harman & Garhammer, 2008). The time in air (TIA) method was used for calculating jump height derived from the force platform (Aragón–Vargas, 2000). Time in air was defined as the period between takeoff and contact after flight (Moir, 2008).

$$\text{TIA jump height} = \tfrac{1}{2}\, g(t / 2)^2, \text{ where } g = 9.81 \text{ m} \cdot \text{sec}^{-2},\ t = \text{time in air}$$

Data were evaluated using SPSS © (Version 16.0). A repeated measures ANOVA was used to assess differences between Vertec jump height and force platform derived jump height. Trial-to-trial reliability analysis of recorded variables used both single (ICC_{single}) and average (ICC_{ave}) measures intra-class correlations. The ICC classifications of Fleiss (1986) (less than 0.4 was poor, between 0.4 and 0.75 was fair to good, and greater than 0.75 was excellent) were used to describe the range of ICC values. A repeated measures ANOVA was also used to assess the differences between the three trials for each instrument. Significant main effects between trials were further analyzed using a Bonferroni-adjusted pairwise comparison. Effect size classifications of Hopkins (2002) (less than 0.04 was trivial, between 0.041 and 0.249 was small, between 0.25 and 0.549 was medium, between 0.55 and 0.799 was large, and greater than 0.80 was very large) were used to interpret the effect sizes. The *a priori* alpha level was set at $p \leq 0.05$ with power and effect size represented by d and η_p^2, respectively. Linear regression analysis was used to develop a prediction equation to estimate Vertec jump height from force platform derived jump height. Assumptions for linearity of statistics were tested and met.

RESULTS: Analysis of the data revealed jump heights derived from the Vertec and force platform were 0.554 ± 0.10 m and 0.402 ± 0.09 m, respectively, which were statistically different ($p < 0.01$). Despite the differences in jump height, reliability was demonstrated by the Vertec (ICC_{single} 0.990; ICC_{ave} 0.997) and force platform (ICC_{single} 0.978; ICC_{ave} 0.992). Repeated measures ANOVA results revealed no significant difference between trials for the force platform ($p = 0.436$; $d = 0.185$; $\eta_p^2 = 0.041$) but a significant difference between trials for the Vertec ($p = 0.031$; $d = 0.660$; $\eta_p^2 = 0.160$). Bonferroni-adjusted post hoc analysis revealed no significant differences between any of the trials for the Vertec. Results of the regression analysis indicated that the force platform was a significant predictor of Vertec jump height ($R^2 = 0.735$, standard error of estimate [SEE] = 0.054 m). Thus, the following regression equation was developed: Vertec jump height = force platform height (1.024) + 0.142 m (Figure 1).

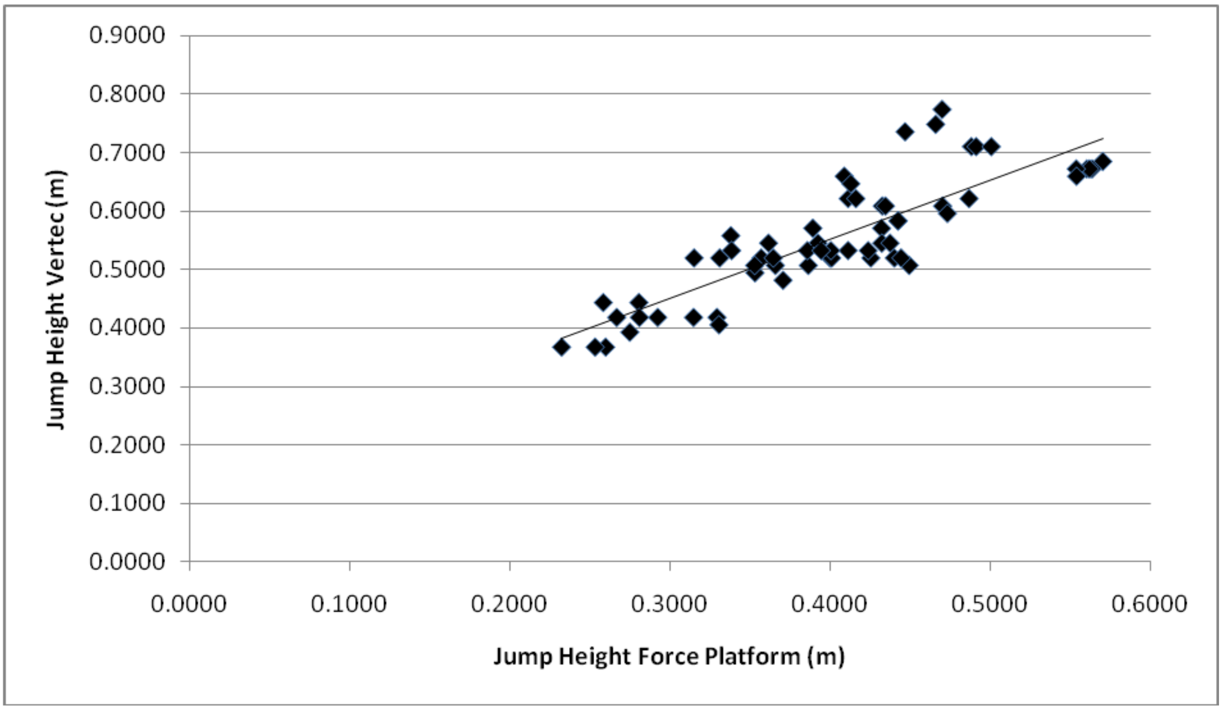

Figure 1. Regression analysis of Vertec and force platform derived jump heights.

DISCUSSION: This is the first study to simultaneously assess differences in Vertec and force platform derived jump heights. Results suggest a 27% higher jump height when assessed by the Vertec, compared to the force platform. The present differences are in contrast to previous reports of lower jump heights obtained from the Vertec compared to contact mats, which use the same calculations to determine jump height as the force platform (Isaacs, 1998; Leard et al., 2007). Details for the procedures and methods for determining Vertec jump height were not explained by Leard and colleagues (2007), thus interpretation of the differences between past and present results remains equivocal. Issacs et al., (1998) however, employed a two arm jump and reach technique and included children, whose motor skills are less developed, which may have influenced the capability of precisely contacting the vanes at the peak of the jump, deflating the Vertec jump height values. General differences between Vertec and force platform jump height assessment could be attributed to the subjects ability to influence initial and final reach height during Vertec testing by manipulating body and or shoulder position to contact the vanes. More specifically, if the subjects body and shoulder position were lower during the initial reach height measurement compared to the maximum jumping reach height, measured jump height would be inflated. Additionally, the number of limbs needed to contact the vanes (1 or 2) for initial and final reach height and the position of the lower body upon landing, could result in jump height discrepancies. Reach height, however, is valuable because of its application to many sports.

Trial-to-trial reliability was excellent for both the Vertec and force platform derived jump heights. These findings are consistent with past research regarding the reliability of force platform derived jump heights (Aragon-Vargas, 2000, Enoksen et al., 2009; Moir, 2008). However, previous research has yet to assess the reliability of Vertec derived jump height (Isaacs, 1998; Leard et al., 2007). Significant differences were found between trials for the Vertec, but post-hoc analysis didn't confirm any differences between trials thus leaving the possibility for type II error. Additionally, the effect size for the differences between Vertec trials was small, thus limiting the possibility for type I error. Nevertheless, the small differences between the trials for the Vertec are likely due to the large (1.27cm) measurement error of the vane spacing. Comparable ICC values were revealed with a device similar to a Vertec, though this device had vanes that were spaced 1.0 cm apart (Young et al., 1997). Results also indicated that Vertec derived jump height can be predicted from force platform data using the regression equation, which allows practitioners to interpret jump heights measured by different instruments.

CONCLUSION: Jump heights derived from a Vertec were 27% higher than heights derived from a force platform. Despite this difference, both instruments provided reliable measurements. Jump height assessment by means of a Vertec is common because of its practicality and external validity. Practitioners will now be able to interpret jump heights derived from force platforms and Vertec's by using the following regression equation: Vertec jump height = force platform height (1.024) + 0.142 m.

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KINETIC QUANTIFICATION OF PLYOMETRIC TAKE OFF, FLIGHT, AND LANDING CHARACTERISTICS

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This study assessed the kinetic characteristics of a variety of plyometric exercises and assessed gender differences therein. Twenty-six men and 23 women performed a variety of plyometric exercises including line hops, 15.24 cm cone hops, squat jumps, tuck jumps, countermovement jumps, loaded countermovement jumps equal to 30% of 1 RM squat, depth jumps normalized to the subjects jump height, and single leg jumps. All plyometric exercises were performed on a force platform. Outcome variables associated with the takeoff, airborne, and landing phase of each plyometric were assessed including the peak ground reaction force during takeoff, time to takeoff, jump height, peak power, peak ground reaction force during landing, and landing rate of force development. A number of differences were found between plyometric exercises.

KEYWORDS: stretch shortening cycle, power, program design, periodization, jump

INTRODUCTION: Explosive exercises such as plyometrics are often used to enhance athletic performance and prevent injury. Results of a meta-analysis demonstrate that plyometric training is effective, though considerable variation exists in the design of plyometric programs employed by researchers (Divillereal et al., 2009). The design of plyometric programs requires an understanding of a variety of variables including exercise intensity (Potach & Chu, 2008).

The intensity of resistance training is easily quantified since most forms of resistance have clearly labeled masses. Resistance training programs are typically progressed using some percentage of an athlete's repetition maximum (RM) or 1RM. Unlike resistance training, plyometric intensity has been defined as the amount of stress placed on involved muscles, connective tissues and joints and is dictated by the type of plyometric exercise that is performed (Potach & Chu, 2008). Based on this definition, it is possible to quantify the intensity of a variety of plyometric exercise based on the kinetic characteristics of the takeoff phase, airborne phase, and landing phase of each exercise.

Previous research has examined ground reaction forces (GRF) and joint reaction forces of a limited number of plyometric exercises such as drop-jumps and pendulum jumps (Fowler & Lees, 1998), unloaded and loaded drop jumps (Tsarouches et al., 1995), drop jumps of varying heights (Raynor & Seng, 1997), and of one-legged and two-legged counter-movement jumps (Van Soest et al., 1985). Research assessing the intensity of a larger number of plyometric exercises is limited to studies quantifying exercise impulse and GRF (Jensen & Ebben, 2007; Jensen et al., 2008), knee joint reaction forces (Jensen & Ebben, 2007), or electromyography (Ebben et al., 2008). Of these studies, some did not assess kinetic variables (Ebben et al., 2008) and those that did used a limited number of subjects (Jensen & Ebben, 2007). The purpose of this study was to quantify plyometric exercise intensity by evaluating kinetic variables associated with the takeoff, airborne, and landing phase of each exercise.

METHODS: Twenty-six men (mean ± SD; age 20.23 ± 1.63 yr; body mass 79.41 ± 9.03 kg) and 23 women (mean ± SD; age 20.39 ± 1.50 yr; body mass 65.35 ± 9.81 kg) athletes served as subjects. The study was approved by the institution's internal review board.
All subjects performed a habituation and testing session. Prior to each session, the subject warmed up and performed dynamic stretching and jumping. During the habituation session, subjects' 5 RM back squat was assessed along with countermovement jump height using a Vertec. Subjects were given instruction, a demonstration, and practiced the correct performance of the plyometric exercises to be tested. The plyometric exercises included line hops (LH), 15.24 cm cone hops (CH), squat jumps (SJ), tuck jumps (TJ), countermovement jumps (CMJ), depth jumps from a box height that was equal to the subjects CMJ height as determined by a Vertec (DJ), loaded countermovement jumps with handheld dumbbells equal to 30% of the subjects estimated 1 RM squat based on their 5RM squat test results (DBJ), and single leg jumps (SLJ). These plyometric exercises were included in this study since they represent a variety of estimated (Potach & Chu, 2008) and researched (Jensen & Ebben, 2007; Ebben et al., 2008) exercise intensities.
During the testing session, subjects performed 3 repetitions of each of the plyometric test exercises in a randomized order with 1 minute of rest between each exercise. The test exercises were assessed with a force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA, USA) which was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA USA) for later analysis. All values were averaged for three trials for each plyometric exercise.
Dependent variables were selected in order to comprehensively evaluate each plyometric exercise including the takeoff phase using vertical ground reaction forces (GRF-T). Data were also assessed for the flight phase using jump height (JH) and power (P). The landing phase of each plyometric exercise was assessed using the landing rate of force development (L-RFD) and landing vertical ground reaction force (GRF-L). These variables were calculated from the force time records of each plyometric exercise consistent with methods previously used (Canavan & Vescovi, 2004; Jensen & Ebben, 2007; Moir, 2008; Raynor & Seng, 1997; Tsarouches et al., 1995; Van Soest et al., 1985). Peak GRF-T was defined as the highest value attained from the force time record for the take off phase of each jump. Jump height and power were calculated based in part on flight time using previously published equations (Moir, 2008). The L-RFD was defined as the first peak of GRF minus the initial GRF upon landing divided by the time to the first peak of GRF minus the time of initial ground reaction force and normalized to one second (Jensen & Ebben, 2007). Peak GRF-L was defined as the highest GRF value attained during the landing phase of the plyometric exercise (Jensen & Ebben, 2007).
The statistical analyses were undertaken with SPSS 17.0. A two way mixed ANOVA with repeated measures for plyometric exercise type was used to evaluate the main effects for plyometric exercise type and the interaction between plyometric exercise type and gender, for GRF-T, JH, P, L-RFD, and GRF-L. Bonferroni adjusted pairwise comparisons were used to identify the specific differences between the plyometric exercises. The trial to trial reliability of each dependent variable was assessed for each plyometric exercise using average measures intraclass correlation coefficient (ICC). In addition, a repeated measures ANOVA was used to confirm that there was no significant difference ($P > 0.05$) between three trials of each plyometric exercise. Assumptions for linearity of statistics were tested and met. An *a priori* alpha level of $P \leq 0.05$ was used with post hoc power and effect size represented by d and η_p^2, respectively.

RESULTS: The analysis of GRF-T revealed significant main effects for plyometric exercise type ($P \leq 0.001$, $\eta_p^2 = 0.60$, $d = 1.00$). Analysis of P showed significant main effects for plyometric exercise type ($P \leq 0.001$, $\eta_p^2 = 0.95$, $d = 1.00$). Analysis of JH showed significant main effects for plyometric exercise type ($P \leq 0.001$, $\eta_p^2 = 0.94$, $d = 1.00$). Analysis of GRF-L showed significant main effects for plyometric exercise type ($P \leq 0.001$, $\eta_p^2 = 0.53$, $d =$

1.00). Finally, analysis of L-RFD showed significant main effects for plyometric exercise type ($P \leq 0.001$, $\eta_p^2 = 0.25$, $d = 1.00$). Results of Bonferroni adjusted pairwise comparisons for each dependent variable are presented in Tables 1 to 5. Intraclass correlation coefficients assessing the trial to trial reliability ranged from 0.34 to 0.99, with most ICC's over 0.80, for the plyometric exercises and dependent variables.

Table 1. Takeoff GRF in Newtons (mean ± SD)

	CH	TJ	CMJ	LH	DBJ	SJ	SLJ
Mean	1311.50[a]	1168.73[b]	1021.33[c]	949.19[d]	864.28[e]	840.55[d]	735.62[c]
SD	353.25	408.33	303.75	284.70	251.90	214.52	238.51

[a]Significantly different (p≤0.001) than all plyometrics except for the TJ
[b]Significantly different (p≤0.001) than all plyometrics except for the CH
[c]Significantly different (p≤0.01) than all other plyometrics
[d]Significantly different (p≤0.05) than all plyometrics except for the DBJ
[e]Significantly different (p≤0.001) than all plyometrics except for the SJ and LH

Table 2. Power in watts (mean ± SD)

	DBJ	DJ	CMJ	TJ	SJ	SLJ	CH	LH
Mean	3946.39[a]	3555.71[b]	3554.81[b]	3497.03[c]	3087.51[a]	2442.88[a]	1989.27[a]	1783.31[a]
SD	983.64	851.75	858.74	804.84	782.85	705.56	569.83	462.85

[a]Significantly different (p<0.001) than all other plyometrics
[b]Significantly different (p<0.05) than DBJ, TJ, SJ, SLJ, CH, LH
[c]Significantly different (p<0.05) than all other plyometrics

Table 3. Jump height in meters (mean ± SD)

	DJ	CMJ	TJ	SJ	DBJ	SLJ	CH	LH
Mean	0.38[a]	0.38[a]	0.37[b]	0.30[c]	0.20[d]	0.20[d]	0.12[c]	0.04[a]
SD	0.08	0.08	0.07	0.07	0.04	0.05	0.04	0.03

[a]Significantly different (p<0.05) than TJ, SJ, DBJ, SLJ, CH, LH
[b]Significantly different (p<0.05) than all other plyometrics
[c]Significantly different (p<0.001) than all other plyometrics
[d]Significantly different (p<0.001) than DJ, CMJ, TJ, SJ, CH, LH

Table 4. Landing rate of force development in $N \cdot m^{-1}$ (mean ± SD)

	DJ	CMJ	DBJ	SJ	TJ	SLJ	CH	LH
Mean	62025.19[a]	56672.28[b]	48922.91[c]	46271.89	44619.92[d]	31007.34[e]	26811.47[e]	17942.71[f]
SD	53699.83	72354.88	39549.69	51560.76	39615.93	20576.40	32067.62	5201.70

[a]Significantly different (p<0.01) than SJ, TJ, SLJ, CH, LH
[b]Significantly different (p<0.01) than SLJ, CH, LH
[c]Significantly different (p<0.001) than SLJ, CH, LH
[d]Significantly different (p<0.01) than DJ, SLJ, CH, LH
[e]Significantly different (p<0.01) than DJ, CMJ, DBJ, SJ, TJ, LH
[f]Significantly different (p<0.05) than all other plyometrics

Table 5. Landing GRF in Newtons (mean ± SD)

	DJ	DBJ	CMJ	TJ	SJ	SLJ	CH	LH
Mean	2589.54[a]	2506.76[b]	2462.35[c]	2285.89[d]	2255.69[d]	1855.47[e]	1458.45[e]	1052.36[e]
SD	995.91	1041.29	1065.21	939.08	929.09	538.67	676.13	453.40

[a]Significantly different (p<0.05) than TJ, SJ, SLJ, CH, LH
[b]Significantly different (p<0.05) than SJ, SLJ, CH, LH
[c]Significantly different (p<0.001) than SLJ, CH, LH
[d]Significantly different (p<0.01) than DJ, SLJ, CH, LH
[e]Significantly different (p<0.05) than all other plyometrics

DISCUSSION/CONCLUSION: A variety of differences in kinetic characteristics between plyometric exercises were found. A number of previous studies assessing plyometric exercises only included one (Raynor & Seng, 1997; Tsarouches et al., 1995) or two plyometric exercises (Fowler & Lees, 1998; Van Soest et al., 1985) in the analysis. Results of the present study can be used to further understand plyometric exercises and prescribe them based on this assessment of the exercise intensity. Practitioners should determine the physical ability they are trying to develop and progress plyometric intensity according to the variables assessed in this study. For example, if developing athletic power is the goal, a plyometric program can be guided by the assessment of power, progressing from low intensity plyometric exercises such as line hops and cone hops, to squat jumps and tuck jumps, to countermovement jumps and depth jumps, and finally to dumbbell jumps and single leg jumps. Similarly, if a practitioner desires to improve and athletes ability to manage the rate and magnitude of landing forces a plyometric program should be progressed from exercises with low L-RFD and GRF-L such as line hops and cone hops, to squat and tuck jumps, to countermovement jumps, dumbbell and depth jumps, and finally single leg jumps. Plyometric exercises can be similarly progress based on other variables assessed in this study.

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THE OSTEOGENIC POTENTIAL OF SUPERMAXIMAL SQUAT LOADS

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This study evaluated the ground reaction force (GRF) and rate of force development (RFD) of the back squat at 3 different loads. Twelve subjects performed the back squat with 80%, 100%, and 120% of their 1 repetition maximum (RM) on a force platform. A two way repeated measures ANOVA revealed significant main effects for GRF for both the eccentric ($p \leq 0.001$) and concentric ($p \leq 0.001$) phases but no interaction between phase and GRF or RFD ($p < 0.05$). No significant main effects were found for RFD for the eccentric ($p = 0.09$) and concentric phases ($p = 0.38$). Post hoc analyses demonstrated that back squats at 120% produced the highest GRF in the eccentric and concentric conditions. Mean RFD was highest, and trending toward significance, during the eccentric phase at 100% of 1 RM condition.

KEYWORDS: resistance training, compressive loading, bone development, strain

INTRODUCTION: The back squat is a common resistance training exercise that is believed to increase athletic ability, strength, and enhance ligament and bone strength (Chandler & Stone, 1991). Exercises that promote osteogenesis are of particular importance to those who are at increased risk of impaired bone health (Jurimae & Jurimae, 2008). Recently, the back squat has been compared to other modes of training including loaded squat jumps, depth jumps, running, and walking in an attempt to quantify the kinetic osteogenetic potential (Ebben et al., 2010). Osteogenic protential of exercise has been proposed to be a function of the magnitude and rate of force development (RFD) (Skerry, 1997) and has been assessed via vertical peak ground reaction forces (GRF) and eccentric RFD, respectively (Ebben et al., 2010).
During dynamic exercises, the external torque is a function of the magnitude of the external load and the moment arm through which the load is expressed. The moment arm changes throughout the range of motion of dynamic exercise (Harman, 2008). For example, during the back squat, the knee joint moment is approximately longest when the knee flexion angle is the greatest (Gullett et al., 2009). Thus, the external torque is greatest during the deeper portion of the squat. This observation, as well as anecdotal clinical observation demonstrate that subjects can handle less load when performing the back squat with greater compared to less depth. Thus, the magnitude of the load, and therefore osteogenic potential, may be limited by the squat depth. The kinetic characteristics of some variations of the back squat exercise have been previously evaluated (Ebben & Jensen, 2002; Gullett et al., 2009). However, no study has compared exercise load, including those that exceed the 1 RM, in order to assess the osteogenic potential.
Therefore, the purpose of this study was to assess the peak vertical GRF, GRF normalized to body weight, and RFD for both the eccentric and concentric phases of the back squat at 80%, 100%, and 120% of the subject's 1 repetition maximum (RM), in order to assess the osteogenic potential of these back squat loading variations.

METHODS: Subjects included 12 men (mean ± SD; age = 22.42 ± 2.54 years; height = 175.05 ± 7.18 cm; body mass = 83.75 ± 15.25 kg; squat 1 RM = 157.10 ± 28.61 kg). Inclusion criteria consisted of men who regularly participated in lower body resistance training. Exclusion criteria included any orthopedic lower limb pathology that restricted athletic functioning, known cardiovascular pathology, or inability to perform exercises with maximal effort. All subjects provided informed consent prior to the study, and the university's internal review board approved the study.

Subjects participated in a habituation and test session. Before each, subjects warmed up for 3 minutes on a cycle ergometer. Subjects also performed 5 slow bodyweight squats, 10 yard forward walking lunge, 10 yard backward walking lunge, 10 yard walking hamstring stretch, 10 yard walking quadriceps stretch, 20 yard skip, and 5 countermovement jumps of increasing intensity. Subjects then rested for 2 minutes.

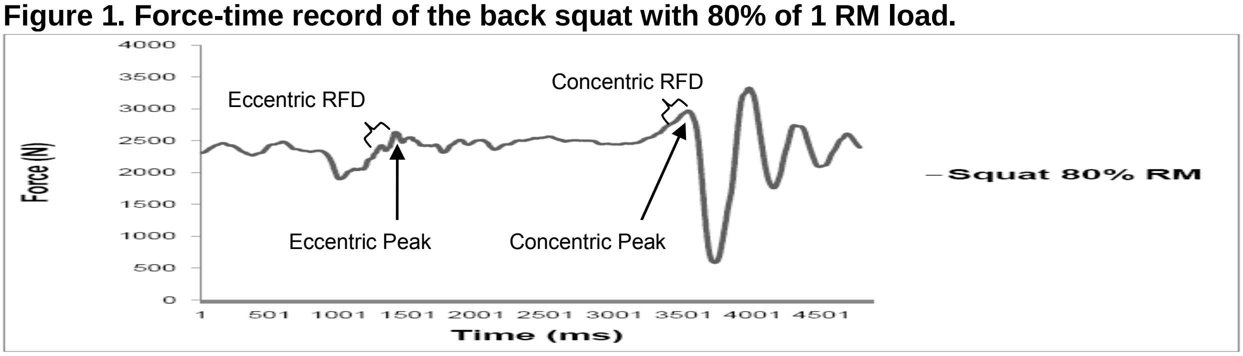
During the habituation session, the subject's age, body mass, height, and history of athletic participation was assessed. Subjects also performed their back squat 5 repetition maximum (RM) down to a knee angle of 90 degrees in order to determine their testing load during the primary testing session. Prior to the 5 RM test, subjects performed 2 sets of 3 reps at approximately 75% and 90% of their self assessed maximum ability. Subjects also perform 2 sets of 1 repetition in the supermaximal (120% of 1 RM) condition at a knee angle of 65 degrees to become familiar with this testing condition.

Subjects then returned for the testing session. During this time, they warmed up using the same warm up protocol as the one used in the habituation session. After 5 minutes of rest, subjects performed 2 sets of 1 repetition of the back squat in the randomly ordered test conditions with 5 minutes rest between sets and exercises. Test conditions included the back squat performed at 80%, 100% and 120% of the subjects estimated 1 RM. The subjects performed the sets of 80% and 100% of the subject's estimated 1 RM loads at approximately 90 degrees of knee flexion. The set at 120% of estimated 1 RM load was performed at 65 degrees of knee flexion, since it was not possible to perform the exercise to 90 degrees of knee flexion with the supermaximal load.

All exercises were performed on a force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA, USA) which was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA USA) for later analysis.

Kinetic data were analyzed for GRF, GRF normalized to body weight, and RFD for both the eccentric and concentric phases of each back squat condition. All values were averaged using 2 test trials. Peak vertical GRF was defined as the highest value attained during the eccentric and concentric phase of each exercise. The RFD was defined as the peak vertical GRF minus the vertical GRF occurring 100 ms prior to the peak vertical GRF and normalized to a second for both the eccentric and concentric phases. Figure 1 shows a sample force-time record for the squat performed in the 80% condition.

Figure 1. Force-time record of the back squat with 80% of 1 RM load.



Data were evaluated with SPSS 18.0 for Windows (Microsoft Corporation, Redmond, WA, USA) using a two way repeated measures ANOVA to determine statistical differences in kinetic data between the exercises and the interaction between GRF and RFD and the eccentric and concentric phase. Significant main effects were further evaluated using Bonferroni adjusted pairwise comparisons. Assumptions for linearity of statistics were tested and met. Statistical power (*d*) and effect size (η^2) are reported, and all data are expressed as means ± SD.

RESULTS: Analysis of GRF demonstrated significant main effects for both the eccentric ($p \leq 0.001$, $\eta^2 = 0.92$, $d = 1.00$) and concentric ($p \leq 0.001$, $\eta^2 = 0.93$, $d = 1.00$) phases, indicating differences in GRF, among the 3 squat loading conditions. There was no significant interaction between GRF and eccentric and concentric phase ($p = 0.11$, $\eta^2 = 0.10$, $d = 0.46$). Analysis of GRF normalized to body weight demonstrated significant main effects for both the eccentric ($p \leq 0.001$, $\eta^2 = 0.91$, $d = 1.00$) and concentric ($p \leq 0.001$, $\eta^2 = 0.91$, $d = 1.00$) phases, indicating differences in GRF normalized to body weight among the 3 squat loading conditions. There was no significant interaction between GRF normalized to body weight and eccentric and concentric phase ($p = 0.17$).
Analysis of RFD demonstrated no significant main effects for the eccentric ($p = 0.09$, $\eta^2 = 0.20$, $d = 0.48$) and concentric ($p = 0.38$, $\eta^2 = 0.08$, $d = 0.20$) phases, indicating no differences in RFD among the 3 back squat load conditions, though the eccentric RFD was approaching significance. There was no significant interaction between GRF and eccentric and concentric phase ($p = 0.33$, $\eta^2 = 0.10$, $d = 0.23$). Significant main effects were further evaluated for both the eccentric and concentric phase and are described in Table 1 and 2. Descriptive RFD data are shown in Table 3.

Table 1. Ground reaction force (GRF) data from the eccentric and concentric phases of 3 squat loading conditions (mean ± SD). (N=12).

	Squat 80% RM	Squat 100% RM	Squat 120% RM
Eccentric GRF (N)	2230.35 ± 316.81*	2625.80 ± 407.55*	2868.30 ± 391.22*
Concentric GRF (N)	2598.16 ± 379.72*	2935.53 ± 390.40*	3306.61 ± 455.20*

*Significantly different than all other squat conditions ($p \leq 0.001$)

Table 2. Ground reaction force (GRF) data normalized to body weight from the eccentric and concentric phases of 3 squat loading conditions (mean ± SD). (N=12).

	Squat 80% RM	Squat 100% RM	Squat 120% RM
Eccentric GRF (BW)	2.77 ± 0.47*	3.26 ± 0.60*	3.57 ± 0.66*
Concentric GRF (BW)	3.23 ± 0.55*	3.65 ± 0.62*	4.12 ± 0.75*

*Significantly different than all other squat conditions ($p \leq 0.001$)

Table 3. Mean rate of force development (RFD) data from the eccentric and concentric phases of 3 squat loading conditions (mean ± SD). (N=12).

	Squat 80% RM	Squat 100% RM	Squat 120% RM
Eccentric RFD ($N \cdot sec^{-1}$)	2844.02 ± 2368.89	3443.69 ± 3056.63	2998.41 ± 1631
Concentric RFD ($N \cdot sec^{-1}$)	1672.63 ± 880.63	1993.69 ± 795.07	1766.28 ± 881.82

DISCUSSION: This study demonstrates that performing the back squat with supermaximal loads of 120% of the estimated 1RM, through 65 degrees range of motion develops higher GRF than performing the back squat with maximal or submaximal loads. Thus, performing this exercise with supermaximal loads may be useful for bone development since the magnitude of the load is believed to be osteogenic (Skerry, 1997). In the present study, a mean increase in back squat load of 20% resulted in a mean increase in eccentric and concentric GRF of 9% and 13%, respectively. While the relationship between squat load and peak GRF may be intuitive, it has not been previously investigated across a loading continuum

or during supermaximal loading conditions. In fact, previous research demonstrated that high load squats offered less GRF than lower load exercises such as jump squats (Ebben et al., 2010). In the present study, squat range of motion, and the likely increase in lengths of the moment arm of the resistance force, likely reduced the possible training load since the moment arm of the resistance is greatest at greater degrees of knee joint flexion (Gullett et al., 2009). Programs designed to optimize osteogenesis should include supramaximal squats as well as loaded squat jumps and depth jumps which have been shown to produce high GRF and RFD (Ebben et al., 2010). From the perspective of sport specificity, it is recognized that training for athletic development may require squatting with greater than 65 degrees of knee flexion for sports that require athletes to function in lower positions (Chandler & Stone, 1991)
Rate of force development was not significantly different between exercise conditions. Significant subject variability exists with respect to the speed of the eccentric and concentric phases despite the fact that all subjects were instructed to perform each as quickly as possible. Nonetheless, the eccentric RFD approached significance during the back squat, with the 100% of the estimated 1 RM demonstrating the highest mean value. The concentric RFD demonstrated a similar mean pattern typified by the highest mean RFD during the 100% of the estimated 1 RM. The supermaximal squat condition at 120% of the estimated 1 RM demonstrated mean RFD values that were slightly greater than the condition at 80% of the estimated 1 RM. These data suggest that the RFD may not be associated with the lightest load. Previous research demonstrated that eccentric RFD was greatest during depth jump landings and progressively lower during jump squats at 30% of the subject's back squat 1 RM and back squat at 5RM load (Ebben et al., 2010).

CONCLUSION: Performing the back squat at supermaximal loads, accomplished with reduced range of motion, results in the highest GRF, and thus should be included in programs designed to promote osteogenesis.

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GENDER DIFFERENCES IN KNEE EXTENSOR AND FLEXOR PERFORMANCE

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The purpose of this study was to evaluate gender differences in peak torque (PT), rate of torque development (RTD), power, and work during isometric (ISOM) knee extension and isokinetic (ISOK) knee extension and flexion. Forty-four university women and men volunteered to perform the test exercises on a computerized dynamometer. Data were reduced with manufacturer software and were analyzed with an independent-samples *t*-test in order to determine gender differences. Results revealed significant differences ($P \leq 0.001$) for ISOM PT and ISOK PT, RTD, power, and work. Gender differences in knee extensor and flexor performance are present, with women demonstrating a range of 68.9 to 76.9% of their male counterparts. Conversely, the rate of force production is not gender specific during ISOM knee extension.

KEYWORDS: strength, torque power, rate of force development, sex differences

INTRODUCTION: Power is defined as the explosive nature of force production (Hakkinen and Komi, 1985). Many sports involve powerful movements, characterized by the generation of force over an acute period of time. Variables utilized to assess such movements include: rate of force development (RFD), rate of torque development (RTD), time to peak torque, and time to takeoff. Previous research has used these variables to test muscular performance within a variety of subject populations. However, normative data and gender differences and similarities are not comprehensively established in athletic populations.

Computerized dynamometry is a reliable and commonly used tool to assess peak torque (PT), RTD, power, and work during isometric (ISOM) knee extension and isokinetic (ISOK) knee extension and flexion exercises (Maffiuletti et al., 2007). Previous research has provided gender and age specific normative values for PT (Freedson et al., 1993; Lindle et al., 1997; Phillips et al., 2000) and prediction equations for PT, power, and work of nonathletic individuals (Neder et al., 1999), but has failed to provide normative values of PT, RTD, power, and work in athletic young adults. Previous studies have used PT to assess gender differences in knee flexion and extension performance, showing that women who suffer anterior cruciate ligament injuries have lower ISOK knee flexion strength compared to men (Meyer et al., 2009). Additionally, normative data for RTD is confounded due to the variety of calculations used.

Gender differences have been demonstrated in some lower body muscular strength tasks (Ebben et al., 2009; Heyward et al., 1986; Mayhew et al., 1994; Mayhew and Salm, 1990; Miller et al., 1993; Sinaki et al., 2001) but not others (Ebben et al., 2007). During vertical jumping, women have been shown to perform at 66.0 to 68.8% of their male counterparts (Mayhew and Salm, 1990; Ebben et al., 2009; Mayhew et al., 1994), whereas during the countermovement jump, women and men have shown similarities in RFD and time to takeoff (Ebben et al., 2007). Therefore, these data from static and dynamic exercises suggest women produce force at a smaller magnitude than men, but their rate of force production is the same. Thus, the purpose of this study was to assess gender differences in PT, RTD, power, and work during ISOM knee extension and ISOK knee extension and flexion.

METHODS: Forty-four university women and men volunteered for this study. Subject data are presented in Table 1.

Table 1. Subject statistics (mean ± SD)

	Women (N=22)	Men (N=22)
Age (years)	20.73 ± 1.28	21.59 ± 1.79
Height (cm)*	167.52 ± 8.94	184.23 ± 9.21
Body mass (kg)*	144.77 ± 13.1	183.86 ± 21.9
High school sports participation (years)	3.92 ± 0.29	3.81 ± 0.25
College sports participation (years)	1.99 ± 1.87	1.76 ± 1.81
Plyometric training participation (days/week)	1.24 ± 1.61	0.96 ± 1.08
Resistance training participation (days/week)	3.16 ± 0.61	3.01 ± 0.79
Aerobic training participation (days/week)	4.01 ± 1.87	3.79 ± 1.93

* Significantly different between women and men ($p \leq 0.05$)

Inclusion criteria consisted of subjects who were 18-27 years old, who participated in NCAA Division I athletics or club or intramural sports, and at least 2 months of biweekly lower body resistance training that which included knee extension and flexion exercises. Exclusion criteria included any history of knee pathology that resulted in functional limitation of the right leg. Subjects provided informed written consent. The study was approved by the institution's internal review board and designed in accordance with the ethical standards of the Helsinki Declaration.

Subjects warmed up and performed 15 seconds of lower body dynamic stretching exercises encompassing each major muscle group. Subjects were then positioned and secured in a dynamometer (System 4, Biodex Inc., Shirley, NY, USA) according to manufacturer specifications. Extraneous movement during the exercise was reduced with the use of straps across the chest, waist, right thigh, and right ankle. The right knee joint was positioned goniometrically at 90 degrees and calibrated with the system software and the mass of the limb was measured. Five seconds of ISOM knee extension were performed at 60 degrees of knee flexion, and ISOK knee extension and flexion were performed concentrically at a speed of 60 degrees · sec. Two test specific warm up sets consisting of three repetitions of each exercise, were performed at 75 and 90%, respectively, of their self-perceived maximum ability.

Five minutes rest was given before the counterbalanced testing sets. Testing consisted of three five second maximal isometric right knee extension and three maximal isokinetic right knee extension and flexion exercises performed at a speed of 60 degrees · sec.

Torque curves for each subject were analyzed using manufacturer's software. Peak torque, RTD, power, and work were calculated for each 5 second isometric sample and each isokinetic knee extension and flexion phase. Rates of torque development were calculated for the first 300 ms of each test exercise and normalized to one second. Absolute torque values were presented in order to make comparisons to previous literature.

All data were analyzed using SPSS 18.0. An independent samples *t*-test was used to evaluate gender differences in PT, RTD, power, and work, as well as differences in subject's demographic data, athletic participation, and training experiences. The *a priori* alpha level was set at $p < 0.05$.

RESULTS: Results reveal a number of gender differences in ISOM and ISOK PT, RTD, power, and work, and are presented in Table 2.

Table 2. Mean PT, RTD, power, and work during ISOM knee extension and ISOK knee extension and flexion for women and men.

	Women (N=22)	Men (N=22)	Ratio	*P*
ISOM Extension PT (Nm)	127.40 ± 18.68	173.61 ± 42.34	73.38	0.001
ISOM Extension RTD (Nm·s^{-1})	282.75 ± 91.28	328.76 ± 89.74	86.00	0.099
ISOK Extension PT (Nm)	118.62 ± 20.21	165.17 ± 32.65	71.82	0.001
ISOK Flexion PT (Nm)	61.33 ± 10.37	86.81 ± 13.05	70.65	0.001
ISOK Extension RTD (Nm·s^{-1})	299.70 ± 72.32	436.52 ± 115.01	68.66	0.001
ISOK Flexion RTD (Nm·s^{-1})	156.46 ± 39.61	237.38 ± 44.70	65.91	0.001
ISOK Extension power (W)	114.27 ± 19.74	161.61 ± 32.89	70.71	0.001
ISOK Flexion power (W)	63.53 ± 9.92	90.86 ± 16.69	69.92	0.001
ISOK Extension work (J)	152.61 ± 25.56	201.22 ± 43.20	75.84	0.001
ISOK Flexion work (J)	88.44 ± 17.66	114.99 ± 23.20	76.91	0.001

Data presented as (mean ± SD); Ratio = represents the ratio of women to men; *P* = *p*-value
ISOM = isometric, ISOK = isokinetic, PT = peak torque, RTD = rate of torque development, W = work

DISCUSSION: This is the first study to investigate gender differences in PT, RTD, power, and work during ISOM knee extension and ISOK knee extension and flexion exercises of an athletic university population. Gender differences were present in ISOM PT and all ISOK variables assessed. However, women and men produce knee extension torque at the same rate during static conditions.

Women in the present study demonstrated 73.4, 71.8, and 70.7% of men's ISOM knee extension and ISOK knee extension and flexion PT, respectively. These gender differences are smaller than those previously demonstrated by women who performed at 63.4, 66.4, and 56.3% the values of men, in the aforementioned variables, respectively (Phillips, et al., 2000). It should be noted that research has revealed no gender differences in lower body force production per unit cross-sectional area of strength-trained individuals (Castro, et al., 1995).

No gender differences were demonstrated in ISOM knee extension RTD, which is consistent with previous research demonstrating no gender differences in RFD and time to takeoff during the countermovement jump (Ebben, et al., 2007). No gender differences were also found in time to peak torque during ISOK knee extension and flexion exercises performed at 180 degrees · sec (De Ste Croix, et al., 2004). However, the current study revealed gender differences in RTD during ISOK knee extension and flexion, with women performing at 68.7 and 65.9% of men, respectively. This gender difference may be explained by women's longer electromechanical delay, which has been thought to result in a slower rate of force production (Bell and Jacobs, 1986). The explanation for the inconsistency in the above absence and presence of gender differences in RTD during static and dynamic powerful movements is unknown.

In the present study, women's performance of ISOK knee extension and flexion, at a speed of 60 degrees· sec , demonstrated power values of 70.7 and 69.9% of men's, respectively. Previous reports of gender differences in power, during ISOK knee extension and flexion at 300 degrees · sec, revealed women results were 61.7 and 55.9% of men's (Neder, et al., 1999). The variability in the gender differences in power is likely due to the speed at which the exercise was performed. Gender mediated differences in neuromotor control may exist thus, potentially negatively affecting women's muscle force production and speed (Karlsson and Jacobs, 1981).

Gender differences in work were also present during ISOK knee extension and flexion, with women performing at 75.8 and 76.9% of their male counterparts, respectively. Previous reports reveal larger gender differences in work, with women performing at 64.9 and 60.1% of men, during the same exercises (Neder, et al., 1999).

CONCLUSION: Results from the present study demonstrate gender differences in ISOM knee extension PT and ISOK knee extension and flexion PT, RTD, power, and work, with women producing 68.9 to 76.9% of the capabilities of men. However, no gender differences were present in ISOM knee extension RTD. Ultimately, these data suggest, in athletic populations, women present smaller gender differences in ISOM knee extension and ISOK knee extension and flexion exercises than previous reports documenting gender differences in nonathletic populations. Future research should evaluate gender differences in strength and power variables of athletic and non-athletic populations.

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JOINT LOADING AT DIFFERENT VARIATIONS OF SQUATS

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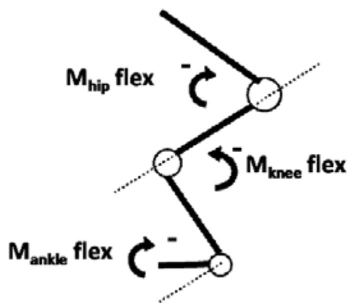
The purpose of this study was to identify the effect of squatting in a common, in a knee-shifted position and in an inclined position (3 cm heel lift) on joint loading. 16 male subjects were tested during squatting with an additional mass of 20 kg. Kinematic and kinetic recordings were performed by two force platforms (AMTI) and a ten infrared camera system (VICON). Inverse dynamics were calculated using a recursive multibody algorithm. Results showed significantly higher ankle dorsiflexion moments as well as higher knee varus moments for the knee-shifted performance. Due to the higher load on the ankle and the knee joint the knee-shifted variation should be avoided in squat training. The inclination of 3 cm does not lead to alterations of the joint moments and therefore does not lead to beneficial effects with respect to joint loading.

KEYWORDS: squats, joint loading, weight training.

INTRODUCTION: Due to the high biomechanical and neuromuscular similarity to sportive movements, such as running and jumping, and to daily living tasks, such as walking, getting up from a chair or step up or down stairs (Flanagan et al., 2003), squats are commonly used as exercise in general fitness training and in rehabilitation programs (Escamilla et al., 1998; Gullet et al., 2008). Also for elderly people the exercise is recommended to maintain their functional ability and, hence, help to provide their physical independence (Flanagan et al., 2003; Salem et al., 2003). Another aspect is the training of young athletes, at which muscle training always has to be considered with respect to functional and adequate joint loading. Squats can be conducted in many different ways. Variations include e.g. different squatting angles (Cotter et al., 2009; Salem & Powers, 2001), lifting additional weight (Cotter et al., 2009) as well as foot posture and stance width (Escamilla et al., 2001). Research focusing on these variations mostly shows little effect of the variations on joint loading. Cotter et al. (2009) prove different joint loading situations while performing squats with additional weight for varying squat angles of the knee. Escamilla et al. (2001) assert that a narrow stance for squat is characterized by lower tibio-femoral compressive forces than for a wide stance. No significant effects of variations in squatting angles (Salem & Powers, 2001) and in foot posture (Escamilla et al., 2001) are observed. Depending on the training goal, different effects might be aimed. In rehabilitation or recreational training the loading on the knee joint should be reduced, while in rehabilitating the patella tendinopathy more loading on the patella tendon seems to enhance the rehabilitation outcome. Good results are therefore reported for squatting on a declined surface, which increases the strain loading in the patella tendon (Kongsgaard et al., 2006; Frohm et al., 2007). Besides the different variations, recommendations are given for the common squat not to let the knee move across the virtual vertical line of the toe to minimize knee joint loading. Escamilla et al. (2009a, 2009b) analyzed cruciate ligament forces (2009a) and the patellofemoral joint force (2009b) at a long wall squat (feet farther from the wall - knee behind vertical line of toe) and short wall squat (feet closer to the wall - knee shifted over vertical line of toes). For the long wall squat higher PCL-forces, but lower patellofemoral joint forces compared to the short wall squat are exhibited, while no research is found though to study the effect of an “incorrect” performance of the common squat in weight training. Given the effect these variations might show, the knee-shift performance also might have an impact on joint loading. In case of any effects, however, they are supposed to be in a similar range as squatting on a declined surface.

Therefore, the aim of this study was to analyze joint moments of three squatting variations representing a common squat, a squat with the knee being shifted over the virtual vertical line of the toe (‘knee-shifted’) and a squat with elevated heels by positioning them on a block of 3 cm.

METHOD: 16 healthy male physically active students (25.1 ± 2.2 years, 183.0 ± 5.8 cm, 80.3 ± 7.6 kg) with no lower extremity injuries participated in this study. Kinematic and kinetic recordings were collected simultaneously by a 10 camera, three-dimensional motion analysis system (VICON, MX camera system, Oxford Metrics Ltd, UK; 200 Hz)[a] and two force platforms (AMTI, model ORG 6; Advanced Mechanical Technology, Watertown, MA, 1000 Hz) embedded in the floor. Reflective markers were placed according to the Vicon Plug In Gait (PIG) markerset[a] including additional markers on the medial epicondyles of the knee and on the barrel. Subjects were instructed to perform 3 different types of squats standing in a natural position. Each foot was standing on one force platform and an additional mass of 20 kg was lifted. The three variations consisted of a common performance (“common” - knees stay behind a virtual vertical line of the toes), a knee anterior shifted performance (“knee-shift” - knee moves across the vertical line of the toes) and a performance, where the subjects position their heels on a wooden block of 3 cm (“block” - block positioned under each heel). Squats were performed to a knee angle of 90°. Tactile feedback was given by a pole, which was positioned horizontally according to the subject’s body height. 8 repetitions were performed for each condition, with each repetition taking 4 seconds and 5 min resting period between each condition.
Sagittal and frontal plane moments were calculated for the hip, knee and ankle using the recursive multibody algorithm MkdTools (Simonidis & Seemann, 2010) and a model based on Zatsiorsky / Seluyanov Parameters (de Leva, 1996). Movements were filtered with a 4 Hz Butterworth filter. Maximum joint moments were identified for each repetition. Ensemble averages of the eight trials were calculated for each parameter. Peak moments are identified in sagittal plane as flexion moments of the hip and knee and as dorsiflexion moment at the ankle. In the frontal plane peak moments are identified as hip abduction moment, knee varus moment and ankle adduction moment (Figure 1) and are presented in Table 1.

a) 

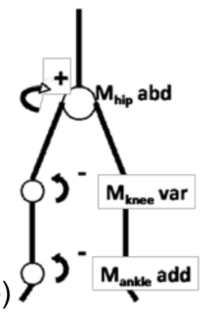
b) 

Figure 2. Peak moments of the ankle dorsiflexion moment (a), the hip adduction moment (b), the knee varus moment (c) and the ankle adduction moment (d).

Only the data of the right leg were considered for statistical analysis. The differences between the variations were statistically tested using an ANOVA with repeated measurements ($p < 0.05$).

RESULTS: Regarding the sagittal plane the ‘knee-shifted’ squat leads to a significant increase of the dorsiflexion moment in the ankle. This is indicated by an increase of 78% compared to the ‘common’ condition and an increase of 104 % compared to the ‘block’ condition (Figure 2.a). No other effects on the joint kinetics have been identified in the sagittal plane. In the frontal plane significant differences are observed at the knee for the ‘knee-shifted’ condition. Compared to the ‘common’ squat the knee varus moment is 127%

increased and compared to the 'block' condition it is increased by 94% (Figure 2.c). The variation of the squat also leads to alterations in the ankle abduction moment, with the least abduction moment for the 'knee-shifted' squat and the highest abduction moment for the 'block' condition (Figure 2.d).

Table 1. Mean maximum moments of hip, knee and ankle joint in sagittal and frontal plane of the right leg; mean (SD)

	common [Nm/BW]	block [Nm/BW]	knee-shift [Nm/BW]
Hip flexion moment	-0.92 (0.25)	-0.90 (0.28)	-0.93 (0.26)
Knee flexion moment	-0.70 (0.23)	-0.76 (0.26)	-0.75 (0.28)
Ankle dorsiflexion moment	-0.32 (0.14)	-0.28 (0.15)	-0.57 (0.2)
Hip abduction moment	0.19 (0.08)	0.17 (0.09)	0.16 (0.09)
Knee varus moment	-0.30 (0.27)	-0.36 (0.27)	-0.69 (0.51)
Ankle adduction moment	-0.03 (0.07)	-0.08 (0.07)	0.00 (0.06)

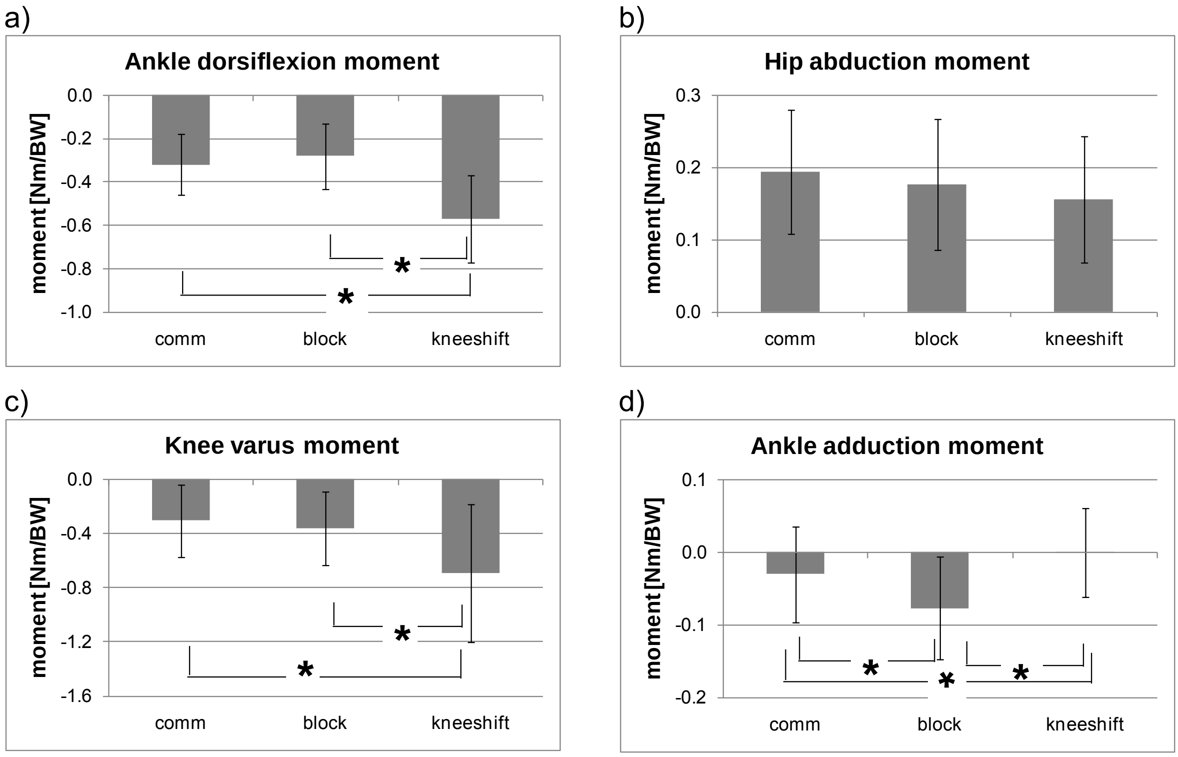


Figure 2. Peak moments of the ankle dorsiflexion moment (a), the hip adduction moment (b), the knee varus moment (c) and the ankle adduction moment (d).

DISCUSSION: The 'knee-shifted' condition with a relatively low additional weight does not affect the knee joint loading in the sagittal plane as one might expect. The only effect in the sagittal plane is a higher dorsiflexion moment at the ankle due to the extended anterior shift of the knee. In the frontal plane the maximum knee varus moment increases significantly. Due to the anterior movement the stabilisation of the knee might be reduced. Furthermore, this positioning leads to higher joint loading. The squat on a block does not lead to alterations in the joint kinetics in this study. Konsgaard et al. (2006) used a declined surface of 25° inclination, while the inclination at this study was ~10°. This inclination seems to be too low to lead to an alteration in the joint kinetics. At the ankle joint significant differences regarding the adduction moment are observed in each condition. Considering the relatively low values of these moments the relevance of these alterations might be neglected. Calculating the knee joint moments is the first step in understanding the forces in the knee joint. By producing a knee flexor moment a quadriceps extensor moment is also needed to hold the subject in equilibrium. Hence, a quadriceps force will be needed to generate this moment, which further has effects on single subcomponents of the knee such as tibiofemoral and patellofemoral compression forces or the forces on the anterior and posterior cruciate ligament (Escamilla, 2001). Therefore this paper only can give a first indication of the effects of variation in squatting technique. For further insight more specific knee models need to be applied to the present data.

CONCLUSION: The ‘knee-shifted’ squat does lead to higher ankle dorsiflexion moments and higher knee varus moments and should, consequently, not be recommended especially for the fitness training in juvenile and elderly athletes. No effect was found in squatting with the heel standing on a block of 3 cm height, despite a higher, but still very low adduction moments at the ankle joint. The chosen heel elevation seems to be too low to lead to signifcant alterations of joint loading. Most likely, however, the stress on the Achilles tendon might be reduced. Alterations in the ankle adduction angle at the three variations exist, but the relevance of these alterations has to be investigated more specifically.

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ECCENTRIC MUSCLE ACTIONS PRODUCE 36% TO 154% LESS ACTIVATION THAN CONCENTRIC MUSCLE ACTIONS

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This study evaluated the differences in eccentric and concentric phase muscle activation of variety of muscles during lower body resistance training exercises. Surface electromyography data (EMG) from 12 subjects was analyzed for the eccentric and concentric phases of the squat, deadlift, step-up, and lunge. Data from the test exercises were averaged for the eccentric and concentric phase for each muscle group to produce a comprehensive measure of activation differences between the eccentric and concentric phases. A paired samples *t*-test revealed differences between eccentric and concentric phase activation for all muscles assessed ($p \leq 0.05$). Results demonstrated that during lower body multi-joint exercises the eccentric phase produced 36% to 154% less muscle activation that the concentric phase.

KEY WORDS: strength training, resistance training, motor unit recruitment, negative work, muscle contraction

INTRODUCTION: Eccentric muscle actions are reported to produce less muscle activation, as measured by electromyography (EMG), than concentric muscle actions under similar loading conditions (Neumann 2002; Zatsiorsky & Kraemer, 2006). These assessments have been typically conducted using isokinetic testing. While differences are known to exist between eccentric and concentric muscle actions, the magnitude of these differences are not clearly understood, especially during multi-joint isotonic resistance training exercises. These differences between eccentric and concentric muscle activation also raises questions about the relative effectiveness of eccentric compared to concentric training and the qualitative differences associated with these muscle actions.

Some evidence comparing eccentric and concentric resistance training shows that eccentric training allows for increased loading and potentially greater adaptation of muscle cross sectional area (Roig et al., 2009). Others report that functional training adaptations are specific to the types of muscle action used in training (Seger & Thorstensson 2005).

Most research examined eccentric compared to concentric differences in muscle activation using isokinetic testing to knee extension (Linnamo et al., 2002; McHugh et al., 2002; Seger & Thorstensson 2005; Westling et al., 1991) or elbow flexion (Fang et al., 2004; Komi et al., 2000; Moritani et al., 1987) exercises. These studies examine the role of muscle length and joint angle (Komi et al., 2000; Linnamo et al., 2002; Moritani et al., 1987), velocity (Westling et al., 1991), loading (Linnamo et al., 2002) cortical activation (Fang et al., 2004), frequency response (McHugh et al., 2002) and assessed preferential activation of fast compared to slow motor units (Komi et al., 2000; Linnamo et al., 2002) between eccentric and concentric muscle actions. These studies also demonstrate that eccentric muscle actions produce greater force than concentric actions (McHugh et al., 2002; Westling et al., 1991) with less muscle activation (Komi et al., 2000; Linnamo et al., 2002; Moritani et al., 1987).

Questions remain about how much difference in muscle activation is present between eccentric and concentric muscle actions and no study has assessed these differences during multi-joint isotonic resistance training exercises. Therefore the purpose of this study was to compare the levels of eccentric and concentric muscle activation associated with a variety of muscles during lower body resistance training exercises for the purpose of quantifying the magnitude of the difference.

METHODS: Subjects included 12 women (mean ± SD; age = 21.00 ± 1.41 years; height = 1.59 ± 0.28 m; body mass = 63.55 ± 6.89 kg) who participated in either NCAA Division I, club, or intramural sports and lower body resistance training. All subjects provided informed consent and the university's internal review board approved the study.

Subjects attended one pre-test habituation session and one testing session. Prior to each, subjects participated in a standardized general and dynamic warm up. During the pre-test habituation session, subjects were familiarized with and performed their 6 repetition maximum (6 RM) for the back squat, deadlift, step-up, and forward lunge. All exercises were performed according to the methods previously described (Earle & Baechle, 2008) with the exception that the step-up began on top of the box, so that all exercises started with the eccentric phase and ended with the concentric phase.

Following the 6 RM testing, subjects were familiarized with 4 maximum voluntary isometric contraction (MVIC) tests for the hamstrings, quadriceps, gluteus medius, and gluteus maximus.

Approximately 1 week after the pre-test habituation session, subjects returned for the testing session. During this session, subjects performed MVICs for the hamstrings, quadriceps, gluteus medius, and gluteus maximus with contractions held for 6 seconds each. Subjects then were tested by performing 2 full range of motion repetitions of their previously determined 6 RM loads, for each of the test exercises. Randomization of the exercises, limited repetitions, and 5 minutes of recovery were provided between MVICs as well as each test exercise.

Surface electromyography (EMG) was used to quantify muscle activation using a fixed shielded cabled, telemetered EMG system (Myomonitor IV, DelSys Inc. Boston, MA, USA). Data were recorded at sample rate of 1024 Hz using bipolar surface electrodes with 1 x 10 mm 99.9% Ag conductors, and an inter-electrode distance of 10 mm. Electrodes were placed on the longitudinal axis of the medial and lateral hamstrings (MH and LH, respectively) the rectus femoris (RF), the vastus lateralis and medialis (VL and ML, respectively), and the gluteus medius and maximus (GMD and GMX, respectively). A common reference electrode was placed on the lateral malleolus. Electrode placement was chosen in order to assess uni-articular and bi-articular knee extensor and flexor muscles, as well as hip abductors and extensors. Additionally, an electric goniometer was placed on the lateral aspect of the right knee in order to distinguish between the eccentric and concentric phases of the test exercises. Skin preparation included shaving, abrasion and cleansing with alcohol. Elastic tape was applied to ensure electrode placement and provide strain relief for the electrode cables. Surface electrodes were connected to an amplifier and streamed continuously through an analog to digital converter (DelSys Inc. Boston, MA, USA) to an IBM-compatible notebook computer.

All data were filtered with a 10-450 Hz band pass filter, saved, and analyzed with the use of software (EMGworks 3.1, DelSys Inc., Boston, MA, USA). The input impedance was 1015 Ohms and the common mode rejection ratio was >80 dB. Raw data were acquired and processed using root mean square (RMS) EMG with a moving window of 125 ms and were analyzed for seconds 2-3 of the MVICs, and for eccentric and concentric phases for the back squat, deadlift, step-up, and forward lunge. Data from the back squat, deadlift, step-up, and forward lunge were averaged for each muscle group for each subject to produce a comprehensive measure of activation for comparison during the eccentric and concentric phases. All RMS EMG values for each muscle were normalized to the average RMS EMG of the 2 trials of the MVIC.

All data were analyzed with SPSS 18.0 using a paired samples *t*-test to determine differences in eccentric and concentric phase RMS EMG for each muscle group. The *a priori* alpha level was set at $P \leq 0.05$ and all data are expressed as means ± SD.

RESULTS: Results reveal that RMS EMG data, expressed as a composite mean and range from all exercises, is significantly different between the eccentric and concentric phases ($p \leq 0.05$). Data are shown in Table 1.

Table 1. Composite mean (range) of muscle activation during the squat, deadlift, step-up and lunge, expressed as a percentage of MVIC for the eccentric and concentric phase (N=12).

MUSCLE	ECCENTRIC PHASE	CONCENTRIC PHASE	DIFFERENCE
LATERAL HAMSTRINGS	0.37 (0.26-0.52)	0.94 (0.59-1.43)	154%
MEDIAL HAMSTRINGS	0.35 (0.23-0.41)	0.66 (0.46-1.01)	80%
RECTUS FEMORIS	0.70 (0.59-0.99)	0.95 (0.78-1.10)	36%
VASTUS MEDIALIS	1.00 (0.75-1.19)	1.68 (1.60-2.15)	68%
VASTUS LATERALIS	0.76 (0.59-0.89)	1.37 (1.17-1.39)	80%
GLUTEUS MEDIUS	0.39 (0.21-0.56)	0.65 (0.36-0.86)	67%
GLUTEUS MAXIMUS	0.89 (0.77-0.97)	1.88 (1.57-2.15)	111%

DISCUSSION: This study is the first to quantify the magnitude of differences in muscle activation between eccentric and concentric phases of several muscle groups during a variety of common, (Earle & Beachle, 2008) multi-joint, lower body resistance training exercises. Results demonstrated that mean eccentric compared to concentric differences are considerable larger that those previously found.

Data from the current study are not presented individually for each of the four exercises and seven muscle groups assessed since this would result in 28 comparisons. Since the purpose of the study was to comprehensively assess the magnitude of the differences in eccentric versus concentric activation as a theoretical phenomenon, this study examined a variety of exercises and muscle groups. Previous published studies assessing differences in muscle activation between eccentric and concentric phases were limited to single joint isokinetic knee extension (Linnamo et al., 2002; McHugh et al., 2002; Seger & Thorstensson 2005; Westling et al., 1991) or elbow flexion (Fang et al., 2004; Komi et al., 2000; Moritani et al., 1987) testing. Other studies examining muscle activation during a variety of lower body resistance training exercises did not compare eccentric and concentric activation (Ebben et al., 2009).

Previous research demonstrated approximately 7-31% lower knee extensor EMG in the eccentric compared to the concentric phase of isokinetic knee extension (Westling et al., 1991; Linnamo et al., 2002). In some cases these values were estimated based on the interpretation of figures since numerical data are not reported (Linnamo et al., 2002). Studies examining eccentric compared to concentric differences during isokinetic elbow extension demonstrated approximately 20-30% less muscle activation (Komi et al., 2000), though in some cases no significant differences were found during the portion of the range of motion that resulted in muscles that were significantly longer than resting length, and up to 97% differences at relatively short muscles lengths. Thus, the findings of the current study demonstrate that dynamic, maximal volitional velocity, full range of motion resistance training exercises result in comparatively larger eccentric to concentric muscle activation differences than typically shown in previous studies that used single joint isokientic testing.

In the present study, multiple muscles and common multi-joint resistance training exercises performed at relatively high training loads and maximal volitional velocity were used to enhance the external validity of the findings. Previous research demonstrated that joint angle, muscle length, and load specific differences are present during single joint isokinetic exercises. The current study assessed all muscles, exercises, and eccentric and concentric phase differences through the full range of motion of each resistance training exercise, consistent with how these exercises are performed in applied settings.

It is possible that the large differences in eccentric to concentric activation found in this study are present in order to provide higher mechanical efficiency as has been proposed (Moritani et al., 1987) which would be particularly valuable for locomotion. Training with eccentric only loading may result in chronic increases in muscle cross sectional area (Roig et al., 2009) and corticol processing is considerably greater during the eccentric phase (Fang et al., 2004) despite significantly lower levels of activation. Future research in this area should assess differences in eccentric to concentric activation between upper and lower body exercises.

CONCLUSION: Results of this study demonstrated that for a variety of lower body multi-joint exercises, concentric activations produced 36% to 154% more muscle activation. This finding helps quantify the nature of the differences between eccentric and concentric muscle actions. Programs designed to stimulate motor unit recruitment should include exercises that include concentric muscle actions and avoid eccentric only training protocols.

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ELECTROMYOGRAPHIC ANALYSIS IN ABDOMINAL MUSLCES DURING CURL-UP EXERCISES

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KEYWORDS: rectus abdominis, external oblique, isometric.

INTRODUCTION: Trunk muscle strength is necessary to maintain the core stability. Several abdominal machine exercises have been investigated through electromyographic (EMG) analysis. Escamilla, et al. (2008) studied numerous abdominal exercises and found that Ab Slide and Torso Track were the most effective exercises in activating abdominal muscles with the relative high EMG activities during abdominal exercise. High internal oblique activities and low rectus abdominis activities in bridging stabilization exercise were also found in EMG analysis (Stevens et al., 2006). One of the most common and convenient ways to strengthen abdominal muscle is the sit-up or curl exercise, which could even be performed at home in healthy populations. However, little research has been conducted to study the curl-up exercise. Therefore, the purpose of this study was to investigate the relative EMG activation levels in rectus abdominis and external oblique during the curl-up exercise. The abdominal strengthening technique with higher EMG activities would be revealed in this study.

METHOD: Twelve healthy male subjects (age 21.0±0.7 years, weight 65.5±7.7 kg, height 175.3±5.0 cm, body mass index 21.1±1.8 kg/m^2) participated in this study. The subjects had not participated in regular strengthening programs. Subjects with surgery history in trunk or lower extremity or any other neuromuscular deficit were excluded from this study.
EMG signals in rectus abdominis and external oblique were measured bilaterally. Biopac EMG system was used in this study. Skin was cleaned with alcohol wrap and four surface EMG electrodes were placed on the abdomen. EMG signals in maximum voluntary contraction (MVC) were first measured for each muscle. Subjects were then asked to perform curl-up exercises with both arms crossed in front of the chest in static or dynamic testing conditions. Subjects had to maintain position at 30 or 60 degrees of trunk flexion for 10 sec in static isometric test. In dynamic test, subjects consecutively performed ten curl-up movements from supine lying to 30, 60, or 90 degrees of trunk flexion and then reversed at the frequency of 40 cycles/min. The feet of the lower extremities were placed on the ground with knee bent. A specific designed goniometer was used to assure the trunk position at a correct angle. There was a two-min resting interval between each testing condition. The testing sequence was random for each subject. Root mean square values of EMG signals were computed and the average EMG data in left and right sides were calculated. EMG data were normalized as a percentage of MVC. One-way ANOVA with repeated measures was performed to compare the EMG activation between different testing conditions (SPSS).

RESULTS: EMG activations during dynamic and static testing conditions were shown in Table 1. For rectus abdominis, dynamic curl-up movements at 0-30, 0-60 and 0-90 degrees in upward phase (43.9%MVC in average) had significantly greater EMG activities than downward phase (30.4%MVC in average) ($p<0.05$). Isometric contraction at 60 degrees (19.9%MVC) significantly had the lowest EMG level in all testing conditions ($p<0.05$).
For external oblique, dynamic curl-up movements at 0-30,0-60 and 0-90 degrees in upward phase and isometric condition at 30 degrees (59.5%MVC in average) had significantly greater EMG activities than dynamic curl-up movement at 0-30,0-60 and 0-90 degrees in downward phase and isometric condition at 60 degrees (40.1%MVC in average) ($p<0.05$). And the dynamic curl-up movement at 0-60 degrees in upward phase (61.3%MVC) had significantly greater EMG levels than dynamic curl-up exercises at 0-30 degrees in upward phase (55.8%MVC).

Table 1. EMG activities (%MVC) in dynamic and static testing conditions

	Dynamic						Static	
Angles	0 - 30		0 - 60		0 - 90		30	60
Phase	Up	Down	Up	Down	Up	Down	Iso	Iso
Rectus Abdominis*	44.1[a] (15.3)	30.2 (13.5)	46.1[a] (16.7)	29.4 (10.0)	44.9[a] (18.0)	31.6 (13.5)	40.6 (24.4)	19.9[c] (13.0)
External Oblique*	55.8[abd] (13.2)	39.0 (14.9)	61.3[ab] (17.2)	38.7 (17.2)	60.9[b] (14.0)	42.5 (14.9)	60.0[b] (21.0)	40.2 (16.9)

Up=upward; Down=downward; Iso=isometric. *$p<0.05$, ANOVA with repeated measures. Multiple comparison ($p<.05$): a) greater than downward phases at 0-30°, 0-60°, and 0-90°; b) greater than Isometric contraction at 60°; c) lesser than all other conditions; d) lesser than upward phase at 0-60°.

DISCUSSION: In addition to machine exercise, this study attempted to use the EMG activation level during curl-up exercise to recognize the optimal strengthening strategy of abdominal muscles. Curl-up exercise in upward phase would be beneficial in increasing muscle load and more recruitment, while statically maintaining posture at 60 degrees of trunk flexion had minimal muscle firing. The findings of this study would be helpful in the implication for muscle strengthening, physical rehabilitation and athletic training.

CONCLUSION: This study identified the most effective method to facilitate abdominal muscle contraction based on an EMG analysis. Curl-up exercise from supine lying to 60 degrees of trunk flexion could produce the highest EMG activities both in rectus abdominis and external oblique, ranging from 46%MVC to 61%MVC.

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EFFECT OF INCREASING VERTICAL CENTRE OF MASS DISPLACEMENT ON THE BIOMECHANICAL STIMULUS OF TRADITIONAL RESISTANCE TRAINING EXERCISES

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This study investigated the effect of systematically increasing vertical COM displacement on the biomechanical stimulus of a traditional resistance training exercise. Fourteen male rugby union athletes performed maximum velocity repetitions of the deadlift to four different final vertical positions with external loads of 20, 40 and 60% 1RM. Significant increases in force, velocity and power were obtained with lifting techniques that resulted in greater vertical COM displacement, although significant interaction effects revealed that improvements were attenuated with heavier loads. These results have applications to strength and conditioning practice, whereby the traditional resistance training exercise stimulus can be augmented without imposing the overly large eccentric musculoskeletal loads characteristic of landing from maximal weighted vertical jumps.

KEYWORDS: Ballistic, power, weight-training.

INTRODUCTION: Performing resistance training with the intention to lift the load as fast as possible is a common training method used among athletic populations. The practice is commonly referred to as 'explosive' resistance training (ERT) and is currently recommended to improve muscular power and athletic performance (ACSM, 2009; Stone 1993). Theoretically, ERT provides an effective training method as both the intent to lift a load as fast as possible and rapid movement velocity have been shown to be important stimuli that elicit high-velocity-specific neuromuscular adaptations (Kawamori 2006). Exercise selection is considered to be an important acute program variable for ERT and the development of muscular power (ACSM, 2009). Customarily, two broad categories of resistance exercises (referred to as traditional and ballistic) are used. However, performing ERT with traditional resistance exercises may not be optimal due to the suggestion that the stimulus is limited by periods of deceleration and reduced force production during the latter stages of the movement (Newton 1996). Instead, researchers generally recommend ERT be performed with ballistic exercises so that force and acceleration can be maintained throughout the movement (Newton 1996; ACSM, 2009). Various ballistic movements (e.g., jump squat and bench throw) are performed by modifying traditional resistance exercises by throwing or jumping with the load at the end of the concentric lifting phase (Newton 1996; Cormie 2007). When these traditional resistance exercises are performed ballistically, there is a significant increase in force, velocity and power production (Newton 1996; Cormie, 2007). However, decelerating these projected loads during ballistic resistance exercises may lead to overuse injuries (Hoffman, 2005). The objective of this study was to quantify the change in biomechanical stimulus as a traditional resistance exercise was gradually modified to a ballistic movement. It was hypothesized that the magnitude of the kinematic and kinetic variables measured would increase with vertical displacement and that significant increases in the mechanical variables could be achieved during augmented vertical displacements that were less than maximum.

METHODS: Fourteen male rugby union athletes (age: 24.1 ± 3.5 yr; stature: 181.1 ± 6.6 cm; mass: 94.1 ± 10.3 kg; 1RM: 171.7 ± 18.2 kg) gave informed consent to participate in this study, which was granted institutional ethical approval. All athletes had extensive resistance training experience and had recently completed an eight week mesocycle where they regularly performed the deadlift movement to the different postures investigated in this study. Data were collected for each subject over two sessions separated by one week. The first session involved 1RM deadlift testing. During the second session subjects performed maximum effort trials with 20, 40 and 60% of their predetermined 1RM. Each load was lifted under four conditions that progressively increased the vertical COM displacement. Condition 1) subjects completed the concentric phase of the movement in an erect standing position with heels in contact with the ground. Condition 2) subjects completed the concentric phase of the movement in an erect standing position with ankles at maximum plantar flexion. Condition 3) subjects completed the concentric phase of the movement by performing a submaximum vertical jump with the external load held at arms' length. Condition 4) subjects completed the concentric phase of the movement by performing a maximum vertical jump with the external load held at arms' length. Two repetitions were performed in each trial to assess reliability.

Trials were performed with a separate piezoelectric force platform (Kistler, Type 9281B Kistler Instruments, Winterthur, Switzerland) under each foot. Displacement, velocity and power data were calculated for the lifter and external load as a single system. This was achieved by incorporating the vertical ground reaction force (VGRF) data and using the principle that the impulse applied to the system equals its change in momentum (Kawamori 2005). Briefly, trials were initiated with subjects standing erect with the load held at arms' length. Changes in vertical velocity of the system COM were calculated by multiplying the net VGRF (VGRF recorded at the force plate minus the weight of the system) by the intersample time period divided by the mass of the system. Instantaneous velocity at the end of each sampling interval was determined by summing the previous changes in vertical velocity to the pre-interval absolute velocity, which was equal to zero at the start of the movement. The position change over each interval was calculated by taking the product of absolute velocity and the intersample time period. Vertical position of the system COM was then obtained by summing the position changes. Instantaneous power was calculated by taking the product of the VGRF and the concurrent vertical velocity of the system.

A general linear model with repeated measures and Bonferroni *post hoc* tests were used to determine significant differences. All statistical analyses were conducted using SPSS Version 15.0, with statistical significance accepted at a level of $p<0.05$

RESULTS: Test-retest reliability for average velocity, peak velocity, average power, peak power, average force, peak force and COM displacement were all high (ICC = 0.92, 0.89, 0.97, 0.94, 0.98, 0.97, and 0.90), respectively. Vertical displacements of the system COM during the ascent phase of the four lifting conditions are presented in Table 1.

Table 1. Vertical displacements of the system COM during the four lifting conditions with 20, 40 and 60% 1RM barbell loads.

Condition	COM Displacement 20% 1RM (cm) Mean ± SD	COM Displacement 40% 1RM (cm) Mean ± SD	COM Displacement 60% 1RM (cm) Mean ± SD
1	49.9 ± 6.4	48.1 ± 6.0	46.5 ± 5.0
2	61.5 ± 9.9	59.8 ± 6.2	51.6 ± 4.7
3	70.3 ± 8.5	66.0 ± 6.6	54.9 ± 4.6
4	89.1 ± 12.0	77.1 ± 9.1	61.1 ± 4.8

Displacement had a significant effect on the biomechanical stimulus of the exercise, as shown by main effects obtained for lifting condition and all variables measured ($p<0.05$). The results demonstrated a positive relationship between vertical displacement and the magnitude of the mechanical variables analyzed (Figure 1). Significant interaction effects of lifting condition and load were obtained for peak velocity, average power, peak power, average force and peak force. The interaction effects reveal that as the external load increased the augmentation of the mechanical stimulus as a result of increased vertical displacement became attenuated, with the greatest attenuations occurring in the maximal jump condition.

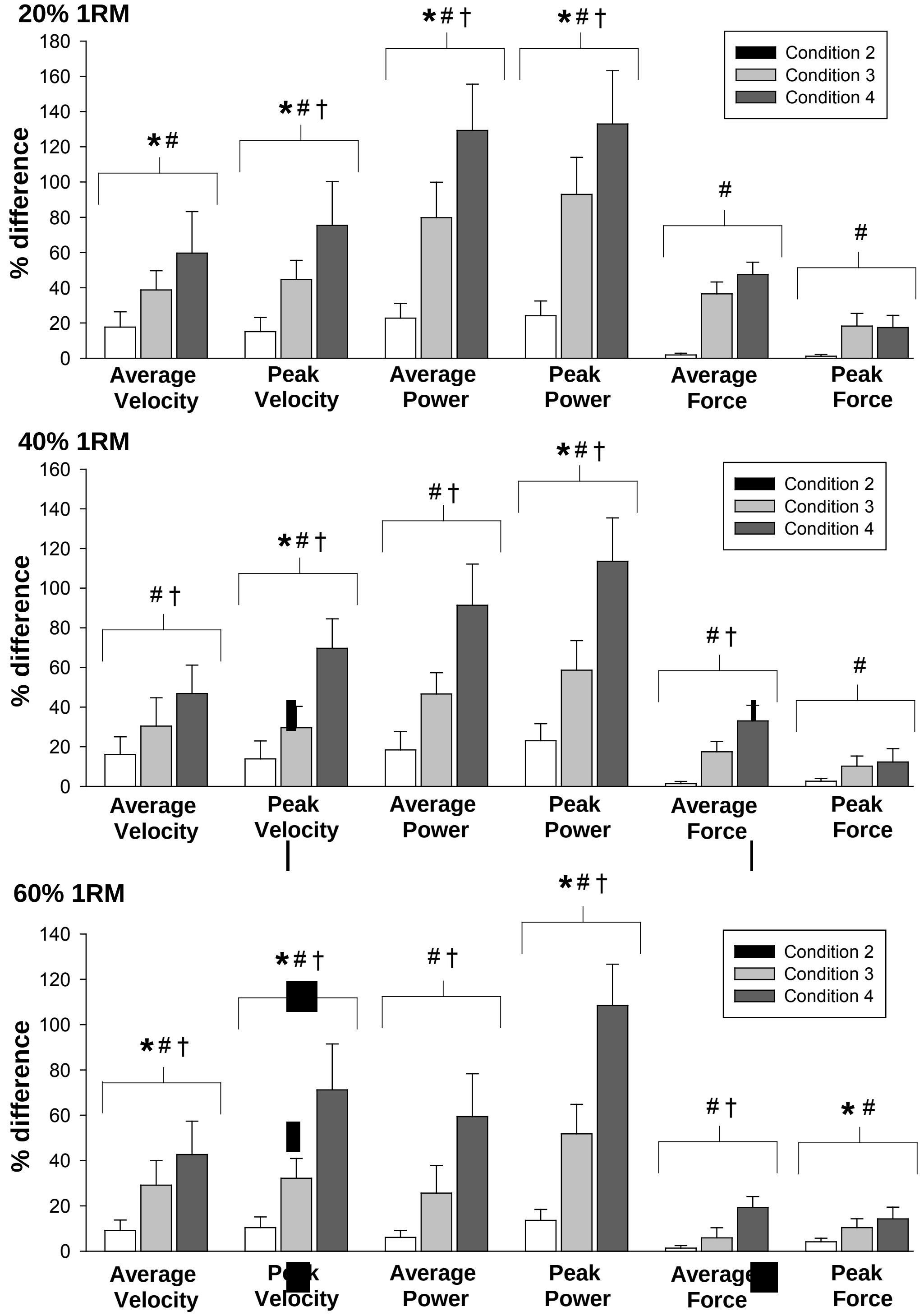

Figure 1. Kinematic and kinetic data for lifting conditions 2, 3 and 4 expressed as a percentage difference relative to lif[redacted]ng condition 1. *Conditions 2, 3 and [redacted] are significantly (p< 0.05) different from condition[redacted]1. #Conditions 3 and 4 are significa[redacted]y (p< 0.05) different from condition 2. †Condition 4 is significantly (p< 0.05) different from condition 3. Error bars represent ± SD.

DISCUSSION: The results of the current investigation show that a positive relationship exists between the vertical displacement of the COM during a resistance training exercise and the magnitude of the force, velocity and power produced. Similar findings have been reported in studies that have compared the mechanical stimulus of the traditional squat and the jump squat (Cormie 2007). By increasing the vertical displacement of the COM the athlete has more time to apply force and change the momentum of the overall system (Frost 2008). In resistance exercises where the body is free to move as a single unit, vertical displacement of the COM will be maximised by jumping with the external load at the end of the movement. Whilst this technique provides the impetus to produce greater amounts of force and power, jumping with an external load may require the athlete to absorb a substantial amount of kinetic energy during the landing phase. Research has shown that large eccentric muscular forces produced to decelerate the system during the landing phase may stimulate physiological adaptations that improve maximum strength (Hoffman, 2005). However, it is also acknowledged that during the landing phase the potential for injury is at its greatest (Hoffman, 2005). When implementing a structured periodization model it may be advantageous to include exercise variations which augment the biomechanical stimulus of traditional resistance training exercises, but do not expose the athlete to the large eccentric loads imposed by maximum weighted jumps. For an exercise such as the deadlift, this study shows that such variations can be obtained by simply plantar flexing the ankles or performing a short jump with the external load held at arms-length at the end of the concentric movement. Future research is warranted to investigate the longitudinal effect of incorporating lifting techniques which alter the vertical COM displacement within a structured periodized model.

CONCLUSIONS:

It is widely accepted that the mechanical stimulus of traditional resistance training exercises are enhanced when an athlete attempts to jump with the load as high as possible. However, the results of the current study reveal that significant increases in force, velocity and power can be obtained by increasing vertical COM displacement by simply plantar flexing the ankles or performing a submaximal vertical jump. This information may prove valuable for strength & conditioning coaches who wish to augment the stimulus of traditional resistance training exercise without imposing the large eccentric musculoskeletal loads that are imposed during the landing phase of weighted maximum jumps.

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LOWER EXTREMITY BIOMECHANICAL ANALYSIS OF A STOP-JUMP TASK WITH DIFFERENT STEP LENGTHS IN THE APPROACH RUN

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This study aimed to assess the kinematics and kinetics during the landing phase of 3 kinds of last step lengths in a stop-jump task to provide further perspectives on lower extremity injuries. Twelve adult males were recruited for the study. A MegaSpeed high-speed camera synchronized with an AMTI force plate was used to record the stop-jump action. Kinetic parameters were calculated using an inverse dynamic method. The results showed that the kinematical characteristics of landing were similar among the different last step lengths during the approach run. The peak vertical ground reaction force and vertical loading rate during landing significantly increased as the step length increased. The peak knee extension moment and proximal tibia anterior shear force did not differ among the 3 stop-jump tasks. These results suggest that during the stop jump task, longer last step lengths during the approach run may increase lower extremity injury.

KEYWORDS: kinematics, kinetics, inverse dynamics

INTRODUCTION: The rate of anterior cruciate ligament (ACL) injury during stop-jump tasks is high (Renstrom et al., 2008). Yu and Garrett (2007) reported that non-contact ACL injuries occur when an anterior shear force generates large forces at the proximal tibia. Previous studies have demonstrated a significant relationship between peak ground reaction forces (GRFs) and knee injury (Williams et al., 2004; Hewett et al., 2005), particularly during ACL loading (Radin et al., 1991; Shelburne et al., 2004). Yu et al. (2006) reported that increasing the peak GRF increased the peak anterior shear force on the proximal tibia during landing in a stop-jump task. The peak posterior GRF during a stop-jump landing is a very important component of the peak proximal tibia anterior shear force. Increasing the knee extension moment by increasing quadriceps muscle activity assists in counteracting the increased knee flexion moment that is created by the larger posterior GRFs experienced during landing (Yu et al., 2006; Yu and Garrett, 2007). Increasing the peak knee extension moment has been shown to increase the peak proximal tibia anterior shear force (Chappell et al., 2002; Yu et al., 2006; Chappell et al., 2007; Sell et al., 2007). Previous investigations are consistent in demonstrating the relationship between kinematics and kinetics—the motion of the hip and the knee in the sagittal plane affects lower extremity loading (Chappell et al., 2002; Yu et al., 2006; Sell et al., 2007; Yu and Garrett, 2007). Unfortunately, these previous studies only focused on the stop-jump task for the subject's preferred step length of the last step during the approach run. However, different last step lengths during the approach run are often used to arrive at a suitable start place for take-off in jumping. Whether different step lengths of the approach run used in the last step during a stop-jump task affect lower extremity loading still is not clear. Thus, the purpose of this study was to compare the kinematics and kinetics during the landing phase of 3 kinds of last step lengths in a stop-jump task.

METHODS AND PROCEDURES: For this study, 12 male, National University of Physical Education students without lower extremity injuries in the 6 months before the experiment were recruited as subjects. The mean age, standing height, and body weight of the subjects were 21.5 ± 0.8 years, 1.74 ± 0.05 m, and 67.3 ± 6.7 kg, respectively. Before the experiment, all subjects were informed of the methods and processes of the study and a signed consent form was obtained. All subjects were blinded to the purpose of this study. A MegaSpeed high-speed camera (120 Hz) was used to record the sagittal plane of the stop-jump task during the landing phase. An AMTI force plate (1200 Hz) was synchronized to calculate the GRFs and center of pressure during jumping. The maximum approach run speed permitted was with 3 steps followed by a full effort stop-jump (a symmetrical two-footed landing and a two-footed takeoff, Figure 1). Each subject performed the stop-jump task with the last step of nature (preferred stop-jump (PSJ)), shorten deliberately (short stop-jump (SSJ)), and lengthen deliberately (long stop-jump (LSJ)) length during the approach run. Three successful stop-jump performances for each step length of the last step during the approach run were collected. The subjects were instructed to land with 2 feet together on the force plate during landing. The highest jumping performance for each stop-jump task was analyzed.

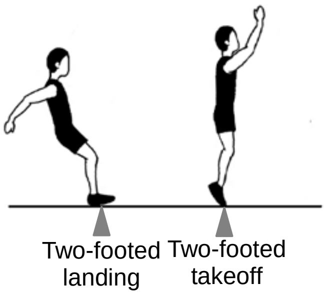

Figure 1. Stop-jump task.

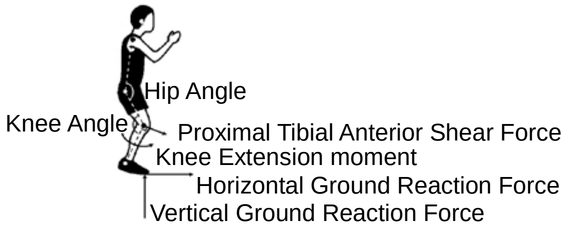

Figure 2. Kinematics and kinetics definition on the landing leg in the sagittal plane.

Sixteen markers were placed on the right and left superior aspects of the scapular acromion process, styloid process of ulna, ulnar styloid, proximal interphalangeal joint of the third finger, greater trochanter, lateral condyle of the tibia, lateral maleolus, and fifth metatarsal, according to Dempster's body segment parameters (Winter, 2005). A reflective marker placed on the edge of the force plate was used to register translational movement. The marker trajectory data were measured and calculated using a Kwon3D motion analysis system and were low-pass filtered with a 4^{th}-order Butterworth filter. All kinematic calculations were performed in the Kwon3D software package. Raw analog data from the force plate were used to calculate the GRF, moments, and center of pressure position by using a KwonGRF system. The inverse dynamic process was used to calculate the net joint reaction forces and net joint moments for the knee (Bresler & Frankel, 1950). Body segment parameters were estimated from the marker data and Dempster's coefficients. All kinetic data were normalized to body weight. The definitions of kinematic and kinetic parameters are shown in Figure 2. The landing phase was defined as the interval between the initial time of landing and the maximum knee flexion angle. The loading rate was defined as the force-to-time ratio, where force is the peak vertical GRF during the landing phase and time is the interval between the initial time of landing and the peak vertical GRF during the landing phase (Winter, 2005). The data were analyzed using the SPSS 14.0 for Windows package program. All data were analyzed using repeated measures one-way analysis of variance (ANOVA) to evaluate whether the median value of the test variable differed significantly among the 3 stop-jump tasks. The significance level was set at $\alpha = 0.05$

RESULTS: The kinematic parameters are presented in Table 1. The last step lengths were significantly differs among the PHJ, SHJ, and LHJ ($P < 0.05$). The hip and knee joint angle at initial foot contact with the ground did not significantly differ among the PHJ, SHJ, and LHJ ($P > 0.05$). The hip and knee joint angle at maximum flexion during landing did not significantly differ among the PHJ, SHJ, and LHJ ($P > 0.05$). The hip and knee joint angular displacement during landing did not significantly differ among the PHJ, SHJ, and LHJ ($P > 0.05$). The hip and knee angular velocity at initial foot contact with the ground did not significantly differ among the PHJ, SHJ, and LHJ ($P > 0.05$). The kinetic parameters are presented in Table 2. During landing, the LHJ had a significantly greater peak horizontal GRF than the PHJ and SHJ ($P < 0.05$). The peak horizontal GRF during landing did not significantly differ between the PHJ and the LHJ ($P > 0.05$). As step length increased, peak vertical GRF during the landing phase increased significantly in the PHJ, SHJ, and LHJ ($P < 0.05$). There was no significant difference in the duration from initial foot-ground contact to the peak vertical GRF among the PHJ, SHJ, and LHJ ($P > 0.05$). As step length increased, vertical loading rate during the landing phase increased significantly in the PHJ, SHJ, and LHJ ($P < 0.05$). During landing, the peak knee extension moment and proximal tibial anterior shear force did not significantly differ among the PHJ, SHJ, and LHJ (P > 0.05).

Table 1. Comparison of lower extremity kinematics (mean [standard deviation]) among 3 landing distances in a stop-jump task

	PHJ	SHJ	LHJ	Post Hoc
Last step length (m)	1.39 (0.14)	1.24 (0.18)	1.93 (0.23)	LHJ>PHJ>SHJ
Hip angle at initial foot contact with ground (deg)	113.65 (12.68)	113.08 (9.41)	111.57 (14.82)	No significant difference
Knee angle at initial foot contact with ground (deg)	141.23 (7.68)	144.19 (7.59)	139.57 (8.40)	No significant difference
Hip angle at maximum flexion during landing (deg)	103.57 (13.35)	104.27 (10.26)	102.45 (11.18)	No significant difference
Knee angle at maximum flexion during landing (deg)	93.45 (10.01)	94.81 (7.53)	94.32 (7.13)	No significant difference
Hip angular displacement during landing (deg)	10.08 (7.69)	8.81 (4.81)	9.12 (11.57)	No significant difference
Knee angular displacement during landing (deg)	47.78 (8.52)	49.37 (6.52)	45.25 (8.06)	No significant difference
Hip angular velocity at initial foot contact with ground (deg/sec)	-3.30 (2.18)	-2.12 (2.57)	-2.54 (1.34)	No significant difference
Knee angular velocity at initial foot contact with ground (deg/sec)	-8.28 (1.35)	-5.92 (4.24)	-8.22 (1.25)	No significant difference

Table 2. Comparison of lower extremity kinetics (mean [standard deviation]) among 3 landing distances in a stop-jump task

	PHJ	SHJ	LHJ	Post Hoc
Peak horizontal GRF during landing (BW)	-0.77 (0.29)	-0.64 (0.19)	-1.38 (0.50)	LHJ>PHJ; LHJ>SHJ
Peak vertical GRF during landing (BW)	1.98 (0.65)	1.50 (0.28)	3.02 (0.60)	LHJ>PHJ>SHJ
Time at which peak vertical GRF occurred following initial foot contact the with ground (sec)	1.24 (0.26)	1.16 (0.20)	1.14 (0.25)	No significant difference
Vertical loading rate (BW/sec)	1.61 (0.48)	1.29 (0.27)	2.76 (0.97)	LHJ>PHJ>SHJ
Peak knee extension moment during landing (Nm/BW)	0.35 (0.09)	0.33 (0.07)	0.39 (0.21)	No significant difference
Peak proximal tibia anterior shear force during landing (BW)	1.10 (0.20)	1.13 (0.22)	1.23 (0.28)	No significant difference

DISCUSSION: The performance of landing in a stop-jump task is important for the overall jumping performance following the landing and for the prevention of lower extremity injuries during landing (Yu et al., 2006). The purpose of this study was to compare the lower extremity loading of a stop-jump using different last step lengths during the approach run. Previous research has demonstrated that hip and knee kinematics in the sagittal plane during a stop-jump landing affect lower extremity loading. Our research shows that there were no significant differences in the hip flexion angle, knee flexion angle, hip flexion angular velocity, and knee flexion angular velocity upon initial foot-ground contact among the PHJ, SHJ, and LHJ. There were also no significant differences in the maximum hip flexion angle, knee flexion angle, and hip and knee angular flexion displacement during landing among the PHJ, SHJ, and LHJ. These results revealed that the kinematical characteristics of landing were similar among the different last step lengths during the approach run. However, the peak horizontal GRF, peak vertical GRF, and vertical loading rate during landing showed a significant increase as step length increased. These results suggest that longer last step lengths during the approach run may raise the risk of lower extremity injury in athletes performing a stop-jump task.

The impact on the lower extremity increases as the peak vertical GRF and loading rate increase (Williams et al., 2004). The results of previous studies showed that a greater vertical GRF and loading rate is associated with knee joint injury (Williams et al., 2004; Hewett et al., 2005), especially in the ACL (Radin et al., 1991; Shelburne et al., 2004). The results of the present study show that peak vertical GRF magnitudes increased significantly as step length increased. However, the time from initial foot-ground contact to the peak vertical GRF was similar among the PHJ, SHJ, and LHJ. According to these results, the loading rate increases significantly as step length increases, thereby increasing the risk of ACL injury.

The results of the present study show that peak posterior horizontal GRF magnitudes were greater in the LHJ. The peak posterior horizontal GRF during the landing of a stop jump may have the effect of muscular moment at the knee, lowering the proximal tibial anterior shear force. Yu et al. (2006) reported that the peak posterior horizontal GRF and peak knee extensor moment are significantly correlated to each other. In addition, the peak posterior horizontal GRF and peak proximal tibial anterior shear force are significantly correlated to each other (Yu et al., 2006). Previous research suggests that peak knee extensor moment and proximal tibial anterior shear force may be a potential risk factor for non-contact ACL injury (Yu et al., 2006; Chappell et al., 2007; Sell et al., 2007). However, this finding was not supported by our study. We found no significant difference in the proximal tibial anterior shear force among the PHJ, SHJ, and LHJ. This finding indicates that posterior horizontal GRF may not be responsible for the different last step lengths of the approach run in the ACL loading of subjects during a stop-jump task. However, muscle electromyographic activity should be studied further for a better understanding of the difference in quadriceps and hamstring muscle activity in the 3 stop-jump tasks.

CONCLUSION: The lower extremity kinematical characteristics of landing were similar among the different last step lengths during the approach run. However, the peak horizontal GRF, peak vertical GRF, and vertical loading rate during landing increased as step length increased, and this may raise the risk of lower extremity injury.

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KINETIC COMPARISON BETWEEN HIGH-IMPACT AND LOW-IMPACT STEP AEROBIC DANCES

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KEYWORDS: Joint force, L- step, Lower limb

INTRODUCTION: Step aerobic dance is one of the most popular aerobic exercises. There are two kinds of aerobic dances, high-impact (HI) and low-impact (LI). High-impact aerobic dance is defined as the exercise involving bouncing, hopping or jumping in which both feet are often taken off the ground. Low-impact aerobic dance is defined as the exercise in which there is always one foot on the ground during the exercise. One of the major causes of sports injury in aerobic dance is overuse injuries of the lower extremity (60%; Francis et al., 1985; Mutoh et al., 1988). This high injury rate in aerobic dance may be due to the repetitive, high joint loads in the lower extremities. Therefore, the purpose of this study was to evaluate the effect of impact level on the joint kinetics of lower limb in step aerobic dance.

METHOD: Eighteen participants with the certification of YMCA Cardio and Step Aerobics Instructor were recruited in this study. A Helen Hayes marker set (19 markers) was bilaterally placed on the selected anatomical landmarks for each participant. VICON612 motion analysis system with a sampling rate of 250 Hz, two force plates (AMTI) with a sampling rate of 1000 Hz and a 15-cm aerobic step (REEBOK) were used in this study. The joint forces of the lower limbs were calculated with the inverse dynamics. Several pairs of aerobic shoes (Touch Aero) were provided for each participant for best-fit of their feet. Each participant was asked to perform the specific-choreographed step aerobic dance for 30 minutes, including warm-up and cool-down. Each participant danced with the same tempo varying from129 to135 beats/min in 30-min music. The step aerobic dance program mainly consisted of low-impact (Figs 1) and high-impact L-steps (Fig 2). Five trials of L-steps were collected and the average was calculated. The joint force was then normalized by body weight (BW) for each participant.

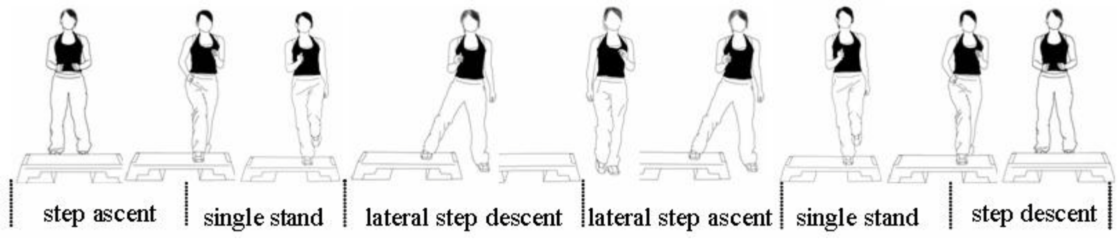

Figure 1. Low-impact L-step movement

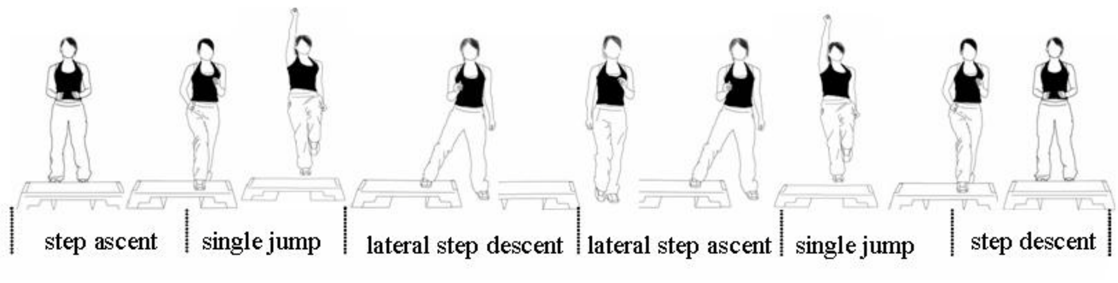

Figure 2. High-impact L-step movement

RESULTS: The joint forces of lower extremity in low-impact and high-impact L-step movements were shown in Table 1. Most of the joint forces in high-impact step aerobic dance were significantly greater than in low-impact step aerobic dance ($p<0.05$), except for four parameters; the anterior hip joint force, the medial knee joint force, and the posterior and medial ankle joint forces. The compression forces showed the greatest magnitude among the six directional forces, corresponding to 1.14 – 1.19 BW in low-impact dance and 1.71 – 1.79 BW in high-impact dance.

Table 1. The joint forces (BW) in the L- step movement (∗*p*<.05, NS: Non-significant).

mean(SD)		Low-impact	High-impact	*p*	Comparison
Hip	Anterior	0.22 (0.05)	0.23 (0.05)	.158	NS
	Posterior	0.40 (0.05)	0.53 (0.06)	.000∗	LI< HI
	Medial	0.05 (0.02)	0.07 (0.03)	.015∗	LI< HI
	Lateral	0.15 (0.03)	0.17 (0.03)	.005∗	LI< HI
	Tensile	0.39 (0.03)	0.43 (0.04)	.000∗	LI< HI
	Compressive	1.14 (0.10)	1.71 (0.20)	.000∗	LI< HI
Knee	Anterior	0.55 (0.09)	0.85 (0.11)	.000∗	LI< HI
	Posterior	0.14 (0.01)	0.15 (0.02)	.000∗	LI< HI
	Medial	0.10 (0.03)	0.10 (0.03)	.060	NS
	Lateral	0.19 (0.01)	0.32 (0.09)	.000∗	LI< HI
	Tensile	0.28 (0.03)	0.30 (0.03)	.000∗	LI< HI
	Compressive	1.19 (0.12)	1.79 (0.20)	.000∗	LI< HI
Ankle	Anterior	0.32 (0.04)	0.55 (0.07)	.000∗	LI< HI
	Posterior	0.09 (0.01)	0.08 (0.01)	.129	NS
	Medial	0.08 (0.04)	0.08 (0.02)	.782	NS
	Lateral	0.29 (0.012)	0.33 (0.13)	.013∗	LI< HI
	Tensile	0.09 (0.01)	0.10 (0.01)	.000∗	LI< HI
	Compressive	1.27 (0.13)	1.95 (0.23)	.000∗	LI< HI

DISCUSSION AND CONCLUSION: Our study illustrated that the joint forces in high-impact step aerobic dance would generate substantial joint loadings in the lower extremities. Compared with the findings in previous studies of stair climbing (Costigan et al., 2002; Riener et al., 2002), the joint compression forces in low-impact step aerobic dance were very close to the joint compressions in stair climbing, while the joint forces in high-impact step aerobic dance were relatively larger, about 1.5 to 2 times of that in stair climbing. Whereas the shear forces in high-impact step aerobic dance were quite similar to the shear forces in stair climbing. This study demonstrated that high-impact step aerobic dance would create a significant axial loading as well as the considerable shear force, especially the anterior knee joint forces (0.85 BW). There would be possibly a risk for the development of joint wear if sustaining long-term high shear force in lower extremity. Our findings suggest that high-impact step aerobic dance might be improper for those who have chronic injury in lower extremity or the novices.

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A PRELIMINARY ELECTROMYOGRAPHIC INVESTIGATION INTO SHOULDER MUSCLE ACTIVITY IN CRICKET SEAM BOWLING

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The aim of this investigation was to describe and compare surface electromyographic activity of shoulder musculature during cricket seam bowling between two elite bowlers with (bowler A) and without (bowler B) shoulder pathology. Activity of seven muscles were recorded at 500 Hz with a digital camera sampling at 210 Hz used to define phases within the movement. Whilst both the duration of the movement and ball velocity were similar between bowlers (bowler A: duration = 0.89 ± 0.04 s, ball velocity = 27.08 ± 1.21 $m.s^{-1}$; bowler B: duration = 0.72 ± 0.02 s, ball velocity = 26.59 ± 1.49 $m.s^{-1}$), variations in muscle activity particularly for biceps brachii and infraspinatus were established. Further research utilising larger sample sizes is required to establish if such variations occur as a consequence of shoulder pathology or if these are due to other contributing factors.

KEYWORDS: EMG, KINETICS, INJURY, CRICKET, ROTATOR CUFF

INTRODUCTION: Shoulder injury prevalence amongst seam bowlers has been reported at 0.9% with consensus amongst researchers that current injury definitions grossly underestimate the true occurrence (Bell-Jenje & Gray, 2005; Ranson & Gregory, 2008). Similar to other overhead athletes, bowlers have been found to exhibit altered joint range and strength associated with internal and external shoulder rotation which may impair the ability of the surrounding musculature to stabilise the joint and prevent migration of the humeral head during deceleration (Aginsky *et al.*, 2004; Giles & Musa, 2008). Whilst altered surface electromyography (sEMG) activity, particularly that of biceps brachii has been associated with shoulder pathology in other sports (Glousman *et al.*, 1988), to date, minimal research has been undertaken to quantify the activation profile of surrounding shoulder musculature during bowling. The aim of this preliminary investigation was to first, describe sEMG activity of the shoulder during the bowling delivery, and second, to compare sEMG activity between two bowlers with and without the presence of shoulder pathology.

METHOD: After gaining university ethical approval, two county medium-fast seam bowlers were recruited and provided informed consent. These bowlers were selected due to displaying similar anthropometric characteristics and both were previously classified by coaching staff as bowling with a semi-open technique. Bowler A (age: 35 years, height: 1.83 m, mass: 76 kg) exhibited a clinical history of injury afflicting his bowling shoulder with an associated change in joint range of motion (internal rotation at 90° abduction: 70°, external rotation at 90° abduction: 125°). In comparison, bowler B (age: 19 years. height: 1.85 m, mass: 67 kg) had no history of injury affecting his bowling shoulder and displayed near symmetrical joint range of motion.
All testing was conducted at the Sussex County Cricket Club Indoor School. sEMG activity of seven muscles (infraspinatus, supraspinatus, anterior deltoid, middle deltoid, posterior deltoid, biceps brachii and triceps brachii) were recorded at 500 Hz using a radio telemetry system (MIE Medical Research Ltd, Leeds, UK). Following skin preparation, AgAgCl surface electrodes were placed in accordance with (Cram *et al.*, 1998), with a maximal voluntary contraction (MVC) recorded against manual resistance for each muscle under investigation. To assist in defining phases of the delivery stride and quantifying ball velocity at release, a digital camera (Casio Exilim EX-FH20 ,Casio, UK) sampling at 210 Hz was positioned parallel to the bowling crease. To enable temporal synchronisation between sEMG and kinematic data, a footswitch was placed in the bowler's footwear to establish front foot contact (FFC).

Following a self-selected warm up, bowlers were instructed to bowl fifteen deliveries at varying lengths to simulate match conditions. After every delivery, bowlers were requested to assess their action to ensure that it was reflective of their normal bowling technique.
The raw sEMG signal was visually appraised to determine its suitability for analysis where two trials for bowler B were excluded from further analysis due to excessive noise. Subsequently all data were imported into a custom program created using Labview 2009 (National Instruments, Austin, USA) where for the purpose of analysing the bowling action, the delivery was divided into four phases. The first phase, pre-delivery stride to back foot contact (PDS to BFC) was signified by the commencement of rotation of the arm during the pre-delivery stride until back foot contact (BFC). The period between BFC and FFC defined the second phase (BFC to FFC), which was followed by the third phase occurring between FFC and the instant of ball release (FFC to BR). The end of the bowling action was defined by the fourth phase, from ball release until follow through, where the bowling arm ceased to rotate (BR to FT). After determination of FFC, the raw signal was rectified and filtered using a low pass filter to create a linear envelope, where it was expressed as a percentage of the MVC value for each muscle. To establish the role of the selected muscles towards both performance and shoulder joint stability throughout the bowling action, muscle activity for each individual muscle, during each phase was analysed in relation to the contribution of average activity and peak muscle amplitude. Average muscle activity was quantified through integration of both MVC and dynamic trials for each muscle using trapezoid rule to provide a standardised measure of the contribution of average activity during each phase in relation to average activity during the entire bowling action. Differences in delivery ball speed and the influence this would impart on muscle activity were accounted for by expressing both peak and average muscle activity values as a percentage of ball velocity.
Statistical analysis was undertaken using SPSS version 17 for windows (SPSS inc., Chicago, USA). Inter-bowler and within-bowler consistency was quantified using the coefficient of variation (CV). To avoid violations of statistical assumptions, comparisons between bowlers were performed using descriptive statistics.

RESULTS AND DISCUSSION: A graphical representation of the mean muscle activity during the delivery stride for each bowler can be seen in Figure 1, where bowler A typically demonstrated greater muscle activity throughout the movement. For all trials the duration of the bowling action and ball velocity were similar between bowlers (bowler A: duration = 0.89 ± 0.04 s, ball velocity = 27.08 ± 1.21 $m.s^{-1}$; bowler B: duration = 0.72 ± 0.02 s, ball velocity = 26.59 ± 1.49 $m.s^{-1}$). The CV values for each muscle between bowlers ranged from 16.4 to 22.4 %, indicating variability between bowler's techniques. Within-bowler CV values for bowler A ranged between 4.6 to 12.1 % and bowler B 11.1 to 17.9 %. Further research utilising more bowlers would be required to ascertain if within and between-bowler variability is reflective of adaptations due to shoulder pathology or if it is indicative of other factors such as bowling experience.

PDS to BFC: Whilst for both bowlers this phase was found to temporally constitute the majority of the bowling action (bowler A: 40 ± 3 %, bowler B: 36 ± 4 %), the contribution of the bowling arm is minimal with the main emphasis during this phase of the movement is for the bowler to successfully convert momentum gained during the run-up from the lower body and trunk to contribute towards ball velocity. This was supported by low average muscle activity for the majority of muscles for with the largest contributors coming from the triceps brachii (bowler A: 37.0 ± 7.0 %, bowler B: 32.8 ± 6.5 %) and posterior deltoid (bowler A: 34.3 ± 5.3 %, bowler B: 34.5 ± 5.4 %), which aid in extension at both the shoulder and elbow as the arm commences clockwise rotation.

BFC to FFC: During this phase the arm continues to rotate clockwise to coincide with being close to horizontal at FFC. This phase of the movement accounted for 25 ± 1 % (bowler A) and 22 ± 2 % (bowler B) of the bowling action, which was characterised by an increase in muscle activity for both bowlers relative to the initial phase. Whilst only minimal changes were associated with bowler A, greater changes in the contribution of average muscle activity for infraspinatus (27.7 ± 3.6 %), supraspinatus (27.9 ± 2.6 %) and posterior deltoid (38.1 ± 5.1 %) were associated with bowler B.

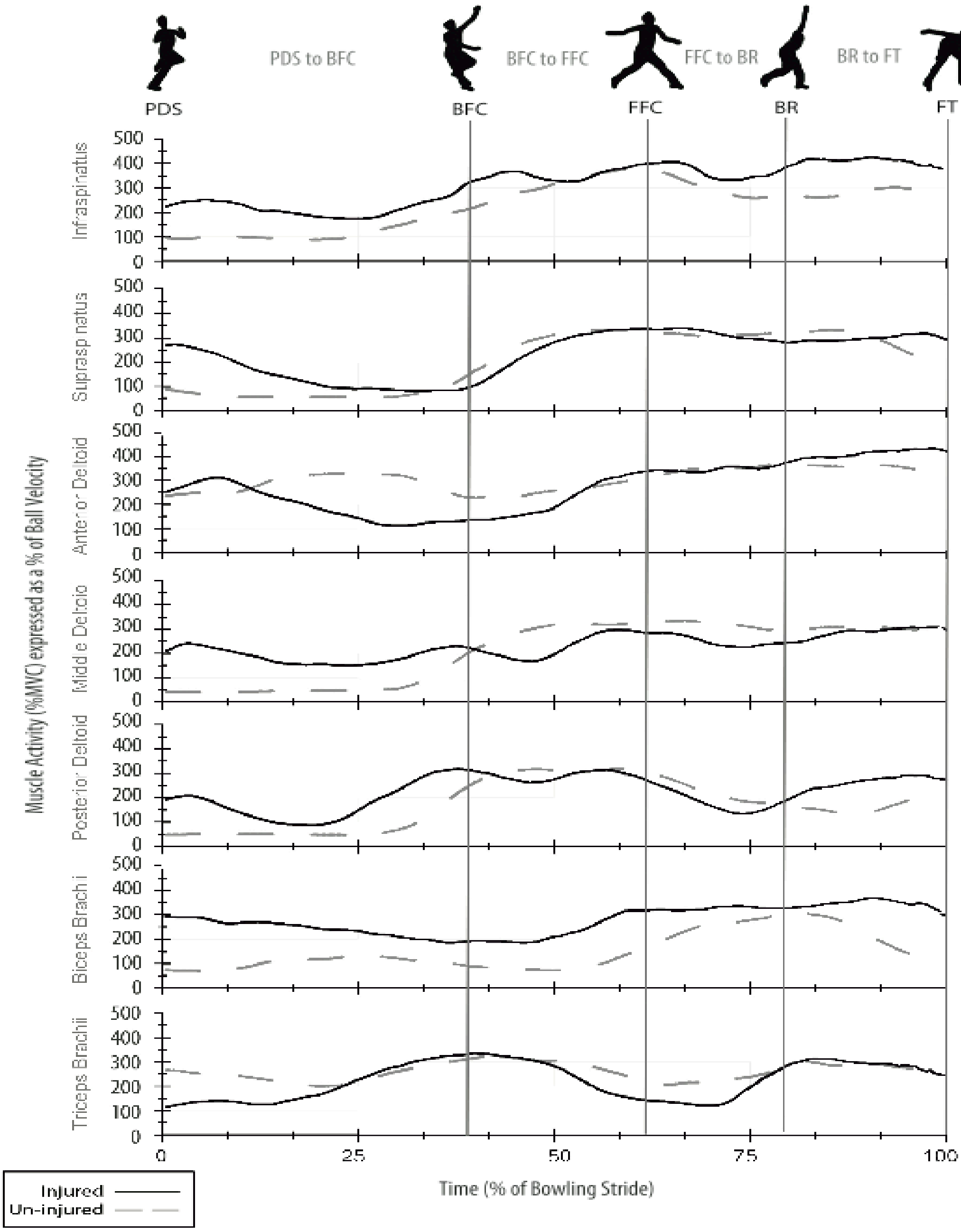

Figure 1. Graphical representation of muscle activity during the delivery stride for bowler A (injured) and bowler B (un-injured)

FFC to BR: This phase of the movement is typified by continued clockwise rotation of the arm in an externally rotated position ending at ball release when the arm is close to the vertical. Although the shortest phase in duration (bowler A: 13 ± 1 %, bowler B: 18 ± 1 %), there were large contributions in peak muscle activity particularly for infraspinatus (bowler A: 424 ± 36 %, bowler B: 408 ± 49 %) and middle deltoid (bowler A: 291 ± 37 %, bowler B: 353 ± 24 %).

BR to FT: During this phase, equating to 21 ± 2 % (bowler A) and 24 ± 2 % (bowler B) of the bowling action, the momentum achieved during the earlier phases causes the arm to continue to rotate through both flexion and adduction at the shoulder towards the final position around the contralateral hip. The focus of the musculature surrounding the shoulder is to control the deceleration of the arm typified by surrounding muscles either eccentrically or concentrically contracting to aid in both arresting the movement and stabilising the joint. Both bowlers demonstrated high contributions of average and peak muscle activity, particularly for infraspinatus (bowler A: peak: 477 ± 54 %, average: 26.6 ± 4.8 %; bowler B: peak: 334 ± 106 %, average: 28.6 ± 4.2 %), anterior deltoid (bowler A: peak: 462 ± 37 %, average: 30.1 ± 4.9 %; bowler B: peak: 403 ± 53 %, average: 28.4 ± 3.9 %) and biceps brachii (bowler A: peak: 397 ± 35 %, average: 25.7 ± 5.2 %; bowler B: peak: 324 ± 49 %, average: 37.0 ± 7.2%).

Throughout the bowling action, variations in muscle activity were observed between bowlers. Whilst there are a multitude of factors that may account for this such as the age and bowling experience of the bowlers analysed, such variation could also occur as a consequence of shoulder pathology. Bowler A demonstrated higher levels of muscle activity particularly for both infraspinatus and biceps brachii, which Glousman *et al.* (1988) postulated may be reflective of a greater reliance on surrounding musculature to maintain joint integrity. The complexity of the joint and the role of surrounding musculature warrant further investigation utilising both kinematic and kinetic techniques to establish the relative contributions of muscle activity towards both bowling performance and joint integrity.

CONCLUSION: Preliminary findings from this study aid in establishing the contribution of shoulder musculature throughout the bowling delivery with variations in muscle activity observed between bowlers with and without the presence of shoulder pathology. Before such findings can be applied, further investigation incorporating larger sample sizes is required first, to substantiate how individualised execution of the bowling action is, and second, to quantify contributions individual muscles have on shoulder joint forces during this dynamic movement.

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Relationship between reaction time and onset of the muscle activation during drop landing

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KEYWORDS: reaction time, the timing of muscle activation, jump landing

INTRODUCTION: Quickness is one of very important factors for athletes in sporting activities. Measuring reaction time reflects how quickly they can move by contracting associated muscles. Reaction time consists of the pre-motor time, as the time from stimulus input to the onset of the muscle activation, and the motor time, as the time from the onset of the muscle activation to the point of body motion begun. In 2004, Demont et al. were reported that a neuromuscular feed forward process as measured by preactivation of the muscle to stabilize joints dynamically during drop landing. This contributed to prevent injuries. Both time of pre-motor and preactivation were the muscle activities that occur before the body motion begins.

The purpose of this study was to clarify the relationship between reaction time and onset of the muscle activity during drop landing.

METHOD: Fifteen healthy female college students (age, 21.9 ± 2.1 years; height, 161.1 ± 5.0 cm; weight, 53.1 ± 5.5 kg) volunteered in this study. All subjects had no previous histories of orthopedic injuries or neurological disorders to their lower extremities. Prior to participation, informed written consent was obtained from each subject according to the university institutional review board policy. To measure reaction time, subjects jumped after the light flashes. It measured five times. Reaction time, time between the jump task occurs after the light flash, was measured using the exclusive measuring instrument REACTION (Takei Co.. Ltd. Japan). The apparatus automatically measured the elapsed time of the task occurrence after the light stimulus (Figure 1). They were instructed to jump immediately after the light flashed. In addition, subjects also completed five bilateral drop landings. Subjects landed from a 0.40m height box to a wooden floor. Before the measurement, a couple of practices were conducted to make subjects understand the task. For all landings, subjects performed in a standardized take-off position; hands were placed on iliac crests. Subjects were not given any special instructions with regards to their landing mechanics to prevent experimenter bias.

The electromyogram (EMG) data during drop landing were collected from 4 muscles; vastus medialis (VM), vastus lateralis (VL), semimembranosus (SM) and biceps femoris (BF). The onset of muscle activities was visually identified as the first point, which is the EMG amplitude of 100ms after the start of landing that exceeded the mean baseline activity by three standard deviations. The definition of the baseline was set as the EMG amplitude of the 100ms after the toe-off from the 0.40m height box. Pearson ‘r’ correlation was conducted between reaction time and the onset of each muscle activities. The statistical significance was set at 0.05.

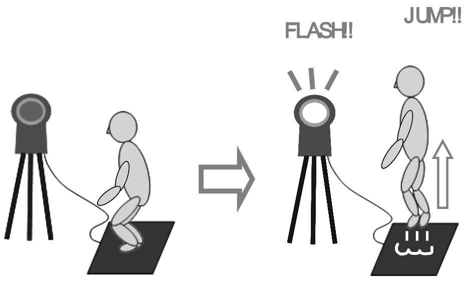

Figure 1. Measuring the reaction time
Subjects jumped after the light flashed.

RESULTS: The mean ± SD of reaction time was 331.8 ± 45.6 ms. The onsets of muscle activities during drop landing were shown in Figure 2. The initial contact occurred 311.3 ms after the toe-off on the box. There were correlations between reaction time and the onsets of SM (r=0.57) and BF (r=0.61) activities ($p<0.05$). However, VM (r=0.38) and VL (r=0.40) activity were no correlations with reaction time.

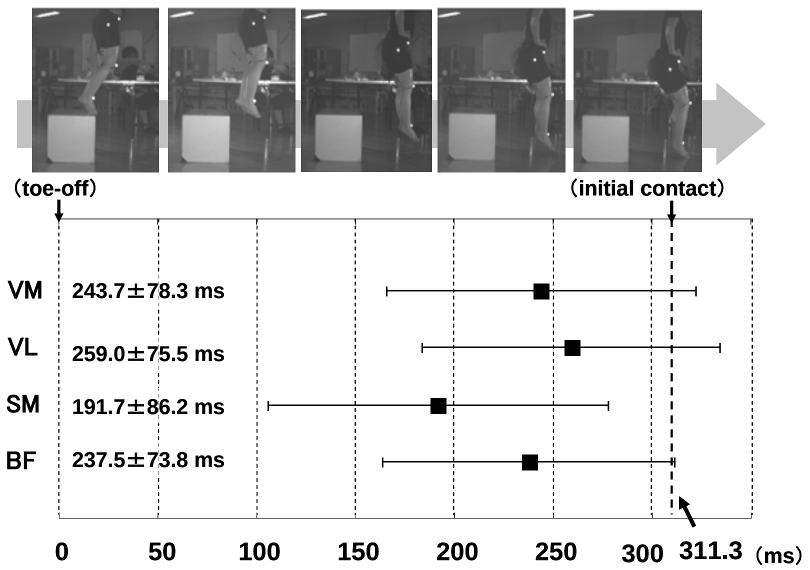

Figure 2. The timing of starting muscle activity during drop landing
Knee extensors: vastus medialis (VM), vastus lateralis (VL).
Knee flexors: semimembranosus (SM) and biceps femoris (BF).

DISCUSSION:

It is reported that the muscle prepares for an impact with activating associated muscles.
An anterior cruciate ligament (ACL) is commonly injured with landing motion. It is suggested that the load on the ACL is reduced by activating hamstring muscles. In 2009, Zebis et al. reported that noninjured female athletes with reduced EMG preactivity of the semitendinosus and increased EMG preactivity of the vatsus lateralis during side cutting were at increased risk of future noncontact ACL rupture. Thus, increasing the knee flexors activities contributed to prevent ACL injuries. In other words, the onset of SM and BF activities faster than those of VM and VL activities would be beneficial to prevent ACL injuries. There were significant correlations between reaction time and the onset of SM and BF activities during drop landing. This study suggests possibility to predict the onset of muscle activities during drop landing by measuring reaction time, though drop landing and jump were different movements.
In the future study, we will clarify the correlation between reaction time and the onset of thigh muscle activities on athletes.

CONCLUSION: This study identified the correlation between reaction time and onset of thigh muscle activities. Because SM and BF activities during drop landing significantly correlated to reaction time, it might be possible to make the onset of SM and BF activities occur faster by reacting to certain stimuli faster.

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DIFFERENCES IN THE FREQUENCY OF MYOELECTRIC ACTIVATION OF LOWER LIMBS BETWEEN SINGLE AND DOUBLE LEG LANDINGS IN MALES

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KEYWORDS: landing tasks, ACL injury, male athletes, EMG, time-frequency domain.

INTRODUCTION: The frequency of myoelectric activation seems to be related to the recruitment of different motor units (Wakeling, 2009). Fast muscle fibers are responsible to the production of higher forces related to the slow fibers (Wakeling, 2009). Therefore, the pattern of activation of some muscles could be associated to the risk of injuries (Bealieau et al., 2008). However, it is not well described whether the muscular activation, in the time-frequency domain, used by males in tasks with different mechanical stresses would truly be related to the strategies of protection of the ACL. The aim of this research was to compare the instantaneous median frequency (IMF) of the EMG signal of lower limbs muscles between different landing tasks in males.

METHODS: Fifteen male athletes (13±1 yr, range 11-14 yr) performed double leg vertical jumps, landing on one leg (SL) or both legs (DL). The project was approved by the Institutional Ethics Committee and all parents signed an informed consent. Before the positioning of the electrodes, the skin was shaved and cleaned with alcohol. The myoelectric signals of rectus femoris (RF), biceps femoris (BF) and hip adductors (HA) were captured (BIOPAC Systems Inc., California) and resolved into their myoelectric intensities in time–frequency space using Choi-Williams transformation technique (Choi, Williams, 1989) (MatLab, Version 7.04, The Mathworks, Inc). The positioning of the electrodes was done according to Cram et al. (1998). EMG signal was filtered with a 4th order Butterworth filter. IMF values were obtained at every 20ms in the window between 100ms before ground contact and 100ms after ground contact, totalizing 10 IMF values. These values were compared between the landings tasks within each phase using Wilcoxon Ranked Test and compared among each phase within each of the landing tasks using Friedman test, with post hoc Dunn's test. Significance level was set at $p<0.05$.

RESULTS: IMF values of RF showed no significant differences when comparing SL and DL within each phase ($p>0.05$). However, when compared within each of the landing tasks RF demonstrated some decrease on IMF values of SL between 40ms and 60ms after ground contact ($p=0.0032$). During the landing phase IMF values of BF tended to be higher in the SL than DL, with statistical significance differences for the 20ms and 40ms interval before ground contact ($p=0.0256$) and 40ms and 60ms after ground contact ($p=0.0181$). Higher IMF values for HA were found during the 40ms to 80ms interval before ground contact compared to 40ms to 80ms after ground contact ($p=0.0011$) for the SL. No statistical differences ($p>0.05$) were observed for DL (Figure 1).

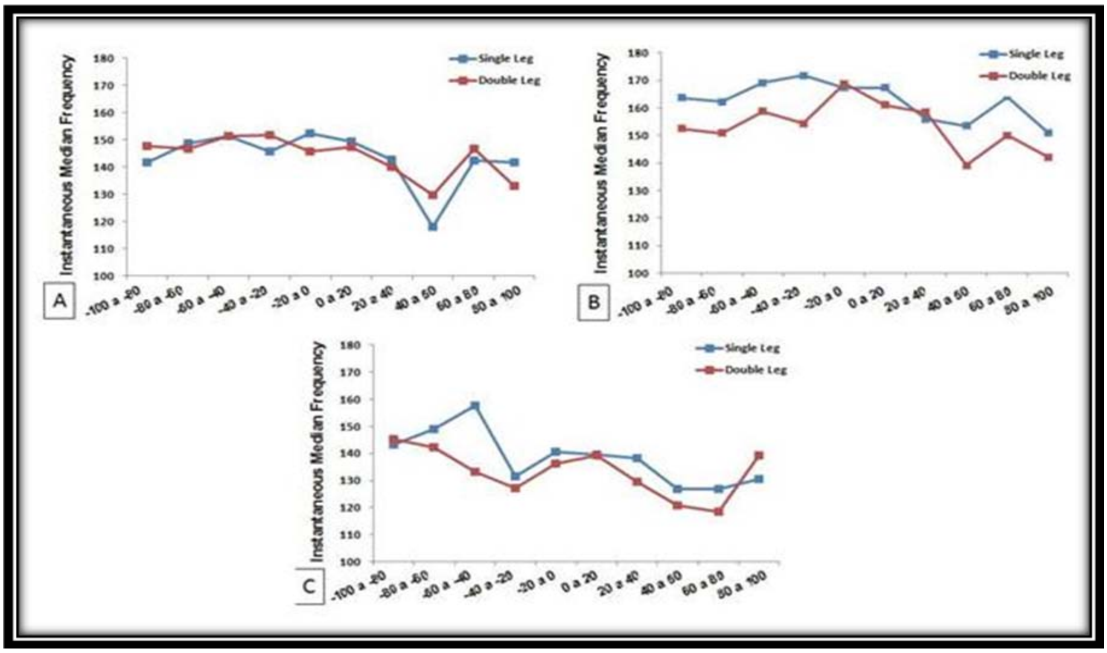

Figure 1. Instantaneous Median Frequency (IMF) values of each of the muscles during the two landing tasks. A: Rectus Femoris; B: Biceps Femoris; C: Hip Adductors.

DISCUSSION: The results suggest a pattern of motor recruitment compatible with strategies to prevent ACL injuries. According to Demorat et al. (2004) RF seems to increase ACL load by increasing anterior tibial shear forces, while BF seems to decrease it. Our results demonstrated that in the SL, RF decreased IMF values and BF increased in the 40ms and 60ms interval, when is believed that ACL injuries occur (Krosshaug et al., 2007). This muscular synergism may increase knee dynamic stability, decreasing ACL tension (Demorat et al., 2004). With the exception of RF, the IMF values for DL tended to be constant, but lower than for SL. The results of HA have demonstrated the strong relation of hip muscles to increase core stability, allowing the activation of distal muscles with safety (Zazulak et al., 2007).

CONCLUSION: The results of this study may provide new insights into the strategies of ACL injury prevention used by male athletes regarding the safety muscular recruitment observed during these both landing tasks.

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EFFECTS OF FEMALE MATURATION ON THE LOWER EXTREMITY BIOMECHANICS DURING THE SIDE-STEP TASK

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KEYWORDS: female, maturation, lower extremity, side step, anterior cruciate ligament.

INTRODUCTION: Anterior cruciate ligament (ACL) injuries are among the most common knee injuries in sports (DeHaven & Lintner, 1986). Female athletes have demonstrated an increased susceptibility to ACL injuries compared to their male counterparts (Yu et al., 2005). The differences in neuromuscular performance during and after puberty may be important contributors to forces on the knee and altered biomechanics could potentially explain the increased risk of ACL injury in females (Quatman et al., 2006). The purpose of this study was to investigate the effects of female maturation on the lower extremity kinematics and kinetics during the side-step task.

METHOD: Twenty-two females participated in this study. The participants were divided into two groups (11 on puberty, age: 12 to 14, height: 152.8±3.9cm, weight: 39.7±5.1kg; 11 post puberty; age: 19 to 21, height: 160.3±3.5cm, weight: 49.8±3.7kg). All participants listed recreational sports, such as basketball and volleyball, as their primary sport. There were no differences in neuromuscular training experience between two groups. The modified Pubertal Matuational Observational Scale (PMOS) was used to classify participants into the 2 maturational categories, on puberty (equivalent to Tanner stages 2 and 3), post puberty (equivalent to Tanner stages 4 and 5), and was assessed during each screening session using parental questionnaires and investigator observations (Quatman et al., 2006). The side-step was performed by planting the right leg on the force platform and followed by left leg from the direction of approach with the approach speed controlled at 3.2±0.3m/s. Three-dimentional videographic (200Hz) and ground reaction force data (2000Hz) were collected performing a side-step task. Statistical analysis consisted of a multivariate test with the level of significance set at P <.05.

RESULTS: The post puberty participants exhibited a significantly greater foot external rotation angle than did on puberty participants (post puberty, 38.79°±14.77°; on puberty, 24.80°±6.94°; P=.048). The post puberty participants demonstrated a significantly greater shank abduction angle than did on puberty participants (post puberty, 12.92°±8.20°; on puberty, 5.37°±5.01°; P=.039). The post puberty participants had significantly greater thigh internal rotation angle than did on puberty participants (post puberty, 25.04°±9.57°; on puberty, 16.86°±4.57°; P=.041). Furthermore, the post puberty participants showed significantly greater knee extension moment than did on puberty participants (post puberty, 34.04Nm/kg±9.01Nm/kg; on puberty, 11.95Nm/kg ±9.80Nm/kg; P=.043).

DISCUSSION: Video analysis of ACL injury during competitive sports play indicates a common body position associated with noncontact ACL injury in which the tibia is externally rotated, the knee is close to full extension, the foot is planted and a deceleration occurs followed by valgus collapse (Hewett et al., 2004). Side-stepping with greater shank abduction, thigh internal rotation angles, and increased knee extension moment in the side-step task may increase the load on the ACL. Female neuromuscular patterns diverge during puberty and show decreased adaptation after puberty. Thus, it appears that the growth and development associated with puberty are related to the neuromuscular and biomechanical factors that underlie the differences in ACL injury risk (Hewett et al., 2004). The results of this study provide that these angles and moments may increase the load on the ACL.

CONCLUSION: The post puberty participants have increased foot external rotation, shank abduction, thigh internal rotaion angles and knee extension moment during the side-step task compared to those of their on puberty counterparts.

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EFFECT OF ANKLE TAPING ON STANDING BALANCE IN THE INDIVIDUALS WITH FUNCTIONAL ANKLE INSTABILITY

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KEYWORDS: ankle sprain, centre of pressure, sports injury.

INTRODUCTION: Ankle sprain is one of the most common sport injuries in athletes. Based on the epidemiologic investigation (Bahr, 1997), the injury rate of ankle sprain could be 54% in volleyball players, indicating that more than half of the volleyball players has been suffering ankle sprain. The rate of recurrent ankle sprain could be as high as 79% in the volleyball players with ankle sprain. Also, ankle sprain is a common sports injury that can cause significant and chronic disability. Functional instability of the ankle has been defined as a tendency for the foot to give away after an ankle sprain. Such instability is a relatively widespread concern following the acute ankle sprain, persisting as a chronic condition long after the apparent signs and symptoms of the original insult have resolved.

Ankle taping has become one of the major interventions in athletic training and is often used for rehabilitation and/or prevention of ankle sprains. Orthotic devices have been shown to effectively modify selected aspects of lower extremity mechanics and improve foot stability during the stance phase of running (Guskiewicz, 1996). Ankle function and muscle coordination after the ankle sprain have been documented (Fu, 2005). However, very little study has been done focusing on the effect of ankle taping on balance control in the individuals with recurrent ankle sprains. Therefore, the purpose of this study was to investigate the effect of ankle taping on the balance ability in the individual with functional ankle instability.

METHOD: Fifteen subjects (10 males and 5 females) participated in this study. Their average age was 21.1±0.96 years. Their average body mass was 65.07±11.41 kg. Their average height was 172.4±8.97 cm. The criteria for subject inclusion were (1) unilateral ankle sprain, (2) at least twice of ankle sprain with first or second degrees in the past three years and (3) no injury within six recent months. Subjects with surgery history in lower extremity or any other neuromuscular deficit would be excluded in this study. They had no pain or uncomfortable symptom in the testing day.

Center of pressure (COP) length was measured with a balance plate (DigiMax posturomed). Subjects were asked to perform one-leg standing in 20 sec. Subjects were asked to perform static standing with slightly knee bended on a balance plate. Arm position was not limited so as to maintain their standing balance as stable as possible. Data were measured in the following testing conditions: 2 sides (healthy and injured), 2 standing surfaces (stable and unstable), 2 visions (eyes-open and eyes-closed) and 2 tapes (taping and no-taping). The healthy and injured ankles were taped by an experienced athletic trainer (1.5” Johnson & Johnson non-elastic white tape). Five-cm displacement in medial-lateral direction was allowed in unstable surface. One trial was performed by each subject in each condition. Vision impaired condition was included in this study because we attempted to see if the standing stability without visual feedback could be enhanced by ankle taping. The COP length of 20-sec data collection was analyzed. Paired t-test was used to compare the difference between taping and no-taping ankles.

RESULTS: COP lengths in different testing conditions are shown in Table 1. For the COP data in anterior-posterior (AP) direction, taping had significantly greater COP lengths than no-taping in the injured ankle in stable and unstable surfaces ($p<0.05$). However, in healthy ankle, taping had significantly lesser COP lengths than no-taping in unstable surface ($p<0.05$). Eyes-open and eyes-closed conditions showed the same findings in AP direction.

The findings in medial-lateral (ML) direction were almost the same with in AP direction, excluding the injured ankle in unstable surface and eyes-closed condition.

Table 1. COP lengths (mm) in different testing conditions

			AP			ML		
Vision	Surface	Side	Taping	No-taping	p	Taping	No-taping	p
Open	Stable	Healthy	29.1	28.6	.721	22.2	21.2	.589
		Injured	35.1	26.2	.015*	26.7	19.9	.002*
	Unstable	Healthy	413.1	847.3	.012*	995.0	1768.6	.015*
		Injured	732.8	340.3	.000*	1968.9	984.1	.004*
Closed	Stable	Healthy	52.5	55.4	.591	38.1	39.6	.662
		Injured	68.9	41.2	.022*	45.7	29.4	.013*
	Unstable	Healthy	539.9	818.6	.008*	974.5	1287.8	.040*
		Injured	856.0	542.7	.014*	1277.7	1037.6	.199

DISCUSSION: Understanding the mechanisms of ankle sprain and balance control is very important in prevention of ankle sprain. Robbins et al. (1995) indicated that ankle taping partially modified impaired proprioception caused by exercise in healthy people. Abian-Vicen et al. (2008) reported that the use of ankle taping had no effect on the jump performance of healthy young subjects. This study demonstrated how the ankle taping influenced on the standing balance in the subjects with functional ankle instability. Based on our findings, ankle taping had no positive influence on standing balance in the ankle with recurrent sprains since tapings showed substantially greater COP lengths than no-tapings almost in all testing conditions for injured ankles. It was implied that taping might deteriorate the ability of balance control in chronic injured ankle. However, taping showed considerably lesser COP lengths than no-taping in healthy ankle in unstable surface. Ankle taping could improve standing stability in healthy stable ankle, especially in unstable surface.

CONCLUSION: This study identified the effect of ankle taping on the standing balance in individuals with recurrent ankle sprains. Ankle taping did not help maintaining better standing balance in the unstable ankle while it could substantially reduce the postural sway in stable ankle in unstable surface. The biomechanical findings of this study would be helpful in athletic training. The athletes could have better performance in sports following the effective prevention of ankle sprain.

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PROPRIOCEPTION OF FOOT AND ANKLE COMPLEX IN YOUNG REGULAR PRACTITIONERS OF WUSHU, TABLE TANNIS AND RUNNING

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The purpose of the study was to examine the possible effects of long-term wushu, table tennis and running on proprioception of the foot and ankle complex in young people. A total of 50 young male students with different exercise habits formed four groups: wushu, table tennis, running, and sedentary control. Kinesthesia of the foot and ankle complex was measured in plantarflexion, dorsiflexion, inversion and eversion at 0.4°/s passive rotation velocity using a custom-made device. The results showed that wushu group had better proprioception than sedentary and table tennis group in dorsi-plantarflexion and better proprioception than table tennis group in in-eversion. Running did not benefit the proprioception of the foot and ankle complex.

KEYWORDS: ankle, kinesthesia, table tennis, Wushu, running

INTRODUCTION： Proprioception is the sensory feedback that contributes to conscious sensation (muscle sense), total posture (postural equilibrium), and segmental posture (joint stability). Research has shown that postural control stability is significantly affected by proprioception in lower limbs (Lord et al., 1991). Colledge et al.(1994) studied the relative contributions to balance of vision, proprioception, and the vestibular system with age by measuring body sway during standing. In four different age groups through 20 to 70 years old, the relative contribution of each sensory input was the same, with proprioception being predominant throughout each age group. Some published research works have found that proprioception can be improved through exercise, especially proprioceptive exercise that requires three actions: the proprioception of the joints, balance capacity, and neuromuscular control (Irrgang & Neri, 2000; Eils & Rosenbaum, 2001). Studies that prove the effects of specific exercise training on proprioception of lower extremities are limited. Xu et al. (2004) and Li et al. (2008) found beneficial effects of long-term Tai Chi practice on the proprioception of ankle and knee joints. Long-term Ice hockey and ballet participants also showed better ankle joint proprioception than their running and sedentary counterparts (Li et al., 2009). As the table tennis and wushu, or the Chinese “Gongfu” such as Tai Chi, are the most popular sports, exercise and physical activity forms in China, it is of interest to know whether long-term, regular practicing track & field and wushu would have positive effects on ankle joint proprioception.

METHODS: Fifty university students were recruited based on their exercise habits. Ten students who regularly practiced wushu for more than three times each week for more than five years formed the wushu (WU) group. Fourteen students who regularly playing table tennis more than three times each week for more than five years formed the table tennis group (TT). Fourteen students with a regular running habit, running for more than three times each week for more than five years, formed the running (RU) group. And 12 students with no regular exercise habit in the past five years served as the sedentary control group (CT). All students were males. The demographic data of the participants are presented in Table 1. All participants were predominantly healthy and they had no history of metabolic, musculoskeletal, or neurological diseases or injuries. An informed consent form was read and signed by each subject prior to participation. This study was approved by the Human Ethics office, the University where the study was conducted.

Table 1. The demographic data of the subjects in each group (Mean ± SD)

Group	Age (years)	Body weight (kg)	Body height (cm)	Body mass index (kg/m^2)
WU (10)	19.93±1.05	64.4±4.86	168.81±2.32*	22.60±1.44
TT (14)	20.06±1.90	69.48±10.50	173.90±2.53	22.93±3.57
RU (14)	21.41±1.33	68.39±1.44	176.95±4.57	21.86±1.77
CT (12)	21.78±2.39	66.86±10.44	173.77±7.54	22.04±2.02

* $P<0.05$ vs.all other groups

The testing was performed in a well-lit and well-ventilated room. The room was sound-attenuated and isolated so as to reduce any auditory or visual interference that might distract the participants. After being measured for body weight and height, each subject individually participated in one session of data collection. Data were collected using the instrumentation and procedures described by previous studies (Xu et al., 2004; Li et al., 2008; Li et al., 2009). The custom-made device was a box with a movable platform that was moved by an electric motor to rotate about a single axis in two directions at a rate of 0.4°/sec. For measurement, each subject was seated on an adjustable chair and his dominant foot was placed on the platform so that the axis of the apparatus coincided with the plantar-dorsiflexion axis or inversion-eversion axis of the ankle joint. The hip, knee, and ankle were each positioned at 90°, respectively. Movement could be stopped at any time with the use of a hand-held switch. The device was also equipped with a hanging scale and a fixed pulley system. Using this system, the investigator standardized and controlled that fifty per cent of each subject's lower extremity weight was rested on the platform. During testing, the subjects' eyes were closed to eliminate visual stimuli from the testing procedure and apparatus. Data collected in each test movement began with the foot placed in a starting position of 0°. The subjects were instructed to concentrate on their foot and to press the hand-switch when they could sense motion and identify the direction of the movement. After performing two practice trials, a formal data collection was conducted. The measurement was arranged in two sessions. In the first session, the motor was randomly engaged to rotate the foot along the ankle sagittal plane for dorsiflexion or plantarflexion at a time interval between 2 and 10 seconds after subject instruction. The researcher recorded the rotation angles of the platform and the direction of movements as passive motion sense. At least 3 angles in each direction were recorded. In the second session, the similar approach was adopted to record the rotation angles of inversion and eversion.
All variables were presented as means and standard deviations. Passive motion sense of foot and ankle complex in two directions of each plane were compared using paired t-test in each group. Because there were no significant differences in the data between the two directions of each plane, the data of the two directions were averaged to present the kinesthesia of foot and ankle complex in the sagittal plane for dorsi-plantarflexion (DP) and in the transverse plane for in-eversion (IV). One-way analysis of variance was used to estimate significant differences among groups. The post hoc Scheffe tests were performed when necessary to isolate the differences and $P \leq 0.05$ was considered statistically significant.

RESULTS: The wushu group showed smaller body height than other groups. There was no difference in body mass index among the groups (Table 1).
The D-P and I-E values of each group are presented in Table 2. Foot and ankle complex kinesthesia significantly differed among the four groups in each plane ($P = 0.001$). The post hoc test showed that in sagittal plane, the control group and table tennis group had higher kinesthesia than wushu group. In the transverse plane, the kinesthesia of the table tennis group was higher than that of the wushu group. No significant difference was found in D-P and I-E among the other groups.

Table 2. foot and ankle complex kinethesia

Kinethesia	CT	TT	RU	WU
D-P	0.47±0.11*	0.47±0.15*	0.37±0.11	0.32±0.08
I-E	0.65±0.15	0.76±0.23**	0.55±0.27	0.51±0.16

Data are mean±SD; *P＜0.05 vs. WU; **P＜0.05 vs. WU

DISCUSSION: The present study provides the evidence that there are significant differences in the perceived passive motion sense between young long-term wushu practitioners, runners, table tennis players and the sedentary people. The long-term wushu practitioners showed significantly better passive motion sense measured in both dorsi-plantarflexion and in-eversion in foot and ankle complex than table tennis players. The long-term wush practitioners also showed significantly better passive motion sense measured in dorsi-plantarflexion than their sedentary counterparts. On the other hand, the long-term running group did not show significant difference in foot and ankle complex kinesthesia as compared with other three groups.

A few publications examined the impacts of exercise on propriocetpion function, specifically comparing the effects of different forms of exercise and the findings are inconsistent. This might be related to the exercise mode and measurement method used in the study. According to Irrgang and Neri's description (2000), propriocpeitive exercises should be composed of three parts: proprioception of joints, balance capacity, and neuromuscular control. A cross sectional study examining the kinesthesia of the knee and ankle joints among three groups of elderly people showed that long-term Tai Chi exercisers had significantly better kinesthesia of the knee and ankle joint than long-term runners or swimmers. Moreover, long-term runners or swimmers could not perform better in perceiving the passive motion of dorsi-plantar flexion of the ankle joint than did a sedentary group (Xu et al., 2004). The present study on young people supported these results. Tai Chi exercise is a series of individual graceful movements in a slow, continuous, circular pattern. The movements of Tai Chi are fluent and consummately precise because specificity of joint angles and body position is of critical importance in accurately and correctly performing each form (Jacobson et al., 1997). Conscious awareness of body position and movement is demanded by the nature of the activity. Such exercise form contains all components that are needed in training and improving proprioception. Li et al. (2008) examined the kinesthesia among young people with four different exercise habits. The results showed that ice hockey and ballet groups perceived significantly better passive motion sense in each inversion and eversion than running group and the sedentary group. No significant difference in the perceived passive motion sense in dorsiflexion, plantar flexion, inversion and eversion was found between running and sedentary groups. The cyclic movement pattern in running and the quick foot maneuvers pattern in table tennis may not contain training effects on ankle joint proprioception.

Some published work could not support the effect of exercise training on proprioception: Schmitt et al. (2005) studied the effects of 5-month ballet training on ankle position sense. Passive angle-replication tests (joint position sense tests) were conducted during the pre- and post-training program. No significant differences in joint position sense were found either in the pre- or post-test of the training program. It is well known that ballet and other dances have a very high demand to proprioception, balance capacity, and neuromuscular control. Ballet training should be considered as a proprioceptive exercise. The possible cause of the undetectable effect of ballet training on proprioception may be related to the sensitivity and reliability of the testing method. Beynnon et al. (2000) compared the accuracy, repeatability, and precision of seven joint position sense techniques and one joint kinesthesia measurement technique in normal subjects with no history of knee injury. They found that joint kinesthesia was more repeatable and precise than each of the joint position sense techniques. They recommended that studies designed to evaluate proprioception should consider using kinesthesia, which should result in increased power and sensitivity to detect significant differences, if they truly exist. In Schmitt et al.'s study (2005), proprioception was examined by measuring joint position sense. The testing method may not have been sensitive enough to detect the proprioception function.

Another factor influencing the impact of exercise on proprioception is the duration of training. Li and co-workers (2008) examined the effects of a 16-week Tai Chi exercise program on the proprioception of the knee and foot and ankle complex in elderly people. The results demonstrated that the significant training effect of kinesthesia was found in the knee joint, but not in the foot and ankle complex. The study by Eils and Rosenbaum (2001) showed that 6-week proprioceptive exercise created gains in the joint position sense in young people. In the study, the subjects trained 20 minutes each day, and the intensity of the 6-week training period was increased by small modifications every 2 weeks. These scientific evidences demonstrated that exercise form, training duration, and age of the participants, as well as the evaluation method used, should be considered in the examination and comparison of the effect of exercise on proprioception.

CONCLUSION: Long-term wushu exercise showed training effects on kinesthesia of the foot and ankle complex in young males. Long-term running and table tennis did not yield training effects on kinesthesia of the ankle and foot complex in the young runners. The results suggest that proprioception of the foot and ankle complex could be improved by exercise in young people.

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MUSCLE ACTIVITY IN THE SUBJECTS WITH FUNCTIONAL INSTABILITY OF THE ANKLE DURING A SINGLE-LEG DROP JUMP

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KEYWORDS: ankle sprain, drop jump, muscle activity.

INTRODUCTION: Ankle sprain is one of the most common injuries experienced sporting participation, and Hertel J (2002) reported it's recurrence rate is very high(47-73 %). Presence of residual pain and functional problems (recurrent complaints of "giving way" or repeated sprain) following inversion ankle sprains are often reported. These symptoms of repeated complaints of "giving way" and/or recurrent sprains have been termed functional instability (FI) of the ankle joint with the report of Freeman, Dean and Hanham (1965). Including the report of Konradsen and Ravn (1991) and Hertsell and Spaulding (1999), There are many studies of muscle functions such as muscle strength, muscle activity, muscle response time of ankle joint evertor in the subjects with FI of the ankle joint. However, a few studies have researched muscle activity in the situation actually occurs ankle sprain such as jump landing on the subjects with FI of ankle joint.
The purpose of this study was to identify differences in ankle joint muscle activity in subjects with FI of the ankle joint during a single-leg drop jump landing.

METHOD: Six male subjects (e; mean ± SD age 23.8 ± 2.99 years; height 177.5 ± 2.52 cm; weight 74.0 ± 6.98 kg) with the subjective compliant of FI, and non-subjective compliant of FI 6 control male subjects (mean ± SD age 23.3 ± 2.88 years; height 172.2 ± 8.06cm; weight 67.5 ± 8.04kg) volunteered to participate in this study. This study is conducted with the approval of the ethics committee of department of physical therapy and occupational therapy sciences (authorization number: 0951). Determination of the subjective complaint FI is made by the Scoring scale by Karlsson et.al (1991), and the total score of 100 points, 80 points or less is judged as FI group.
Subjects performed single leg drop jump on stable surface, and unstable surface created by the balance pad (Airex AlcanAirex, Switzerland). Subjects who remained barefoot during testing, were first introduced to the required jumping technique. Subjects standing on a 40 cm high platform in front of the each surface with the test leg relaxed and non-weightbearing. The subjects then used to opposite limb to propel himself from the platform and landing on the each surface with test leg. Subjects permitted a period of practice in the jumping technique prior to testing. Each subject performed 3 single-leg drop jumps onto the surface of the two conditions. A jump was discarded if the subject required any corrections following landing such as touching the floor with the opposite side foot.
A total of five muscles; tibialis anterior (TA), peroneus longus (PL), gastrocnemius (GAS), soleus (SO), tibialis posterior (TP) were recorded in each group by using the surface EMG (Personal-EMG, Oisaka electronic device Co., Japan). Data relating to the period from initial contact (IC) to 200 ms post-IC were extracted 3 EMG records for each muscle, and these were converted into root mean square (RMS). To compare the muscle activity, data were expressed in percentage of isometric maximal voluntary contractions (%MVC) for each muscle. Subjects during single-leg drop jump were captured by using two high-speed camera units from the front and side of them, and made them synchronized with the EMG to determine the moment of IC for each subject.
Statistical analysis was performed by means of excel add-in software (Statcel 2, OMS-publising, Japan). Differences in FI and control group %EMG in each surface in terms of each muscle were tested for statistical significance using Mann-Whitney U test. Differences in stable surface and unstable surface %EMG in each group in terms of each muscle were

tested for statistical significance using Wilcoxon signed rank test. The level of significance for all analyses was set at $P<0.05$.

RESULTS: The mean ± SD of %MVC activity in each group, each surface, each muscle during the period from IC to 200 ms post-IC are shown in Table 1. FI subjects showed a significant decrease in unstable surface PL %MVC from stable surface ($P<0.05$). No significant differences were noted between the groups each surface in terms of TA, PL, GAS, SO, TP. There were no significant differences between the surfaces each group in terms of TA, GAS, SO, TP.

Table 1. %MVC activity during the period from Initial Contact to 200-ms post-Initial Contact

	FI Group(%)		Control Group(%)	
Variable	Stable surface	Unstable surface	Stable surface	Unstable surface
TA	57.5±26.2	47.1±25.9	55.8±16.8	43.7±11.3
PL	76.2±20.5	59.2±9.50*	62.2±25.9	56.3±30.7
GAS	56.5±28.3	65.6±32.0	51.1±31.9	55.2±32.7
SO	89.4±33.8	80.7±19.7	83.9±43.9	83.5±25.0
TP	97.5±22.0	87.3±24.5	86.4±26.6	86.4±21.3

Values are means ± SD.
*Significant difference from stable surface in FI group (P<0.05).
FI, functional instability.

DISCUSSION: Reimann, Myers and Lephart (2003) reported the importance of ankle joint for shingle-leg stabilization on the unstable multiaxial surface (5). In this study, PL muscle activity was significantly decreased on the unstable surface compared to the muscle activity on the stable surface in FI group ($p<0.05$). The main function of PL muscle is to control the amount of inversion occurring at the ankle joint. The deficit of muscle activity of eversion ankle joint would increase the risk of occurrence of ankle inversion sprain, or of recurrence. Based on the result of this study, it is necessary to intervene to improve the PL muscle activity, using the proper exercise.

CONCLUSION: We have studied that muscle activity in subjects with FI of the ankle joint during a single-leg drop jump. This study has shown that subjects with FI have significantly deficit PL muscle activity on the unstable surface compared to the stable surface.

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THE KNEE JOINT MOMENT AND POWER DURING BALLET'S SIMPLE GROUND ÉCHAPPÉ- COMPARISON OF DIFFERENTIAL PHYSICAL CONDITION IN DANCERS WITH AND WITHOUT KNEE PAIN

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The purpose of this study was to investigate the differential effect of physical condition on the impact force and the knee joint kinematics and kinetics of the dominant leg during ballet's simple ground échappé in ballet dancers with and without anterior knee pain. Ten Eagle cameras (200 Hz) and two AMTI force platforms (2000 Hz) were synchronized to collect the data. The study showed ballet dancers exerted greater knee extensor moment and power generation in hard physical condition than in easy condition during simple ground échappé. Moreover, ballet dancers with knee pain tended to protect their knees with greater knee power absorption during simple ground échappé compared to dancers without knee pain.

KEYWORDS: ballet dance, inverse dynamic, anterior knee pain.

INTRODUCTION: Ballet dance is physically demanding and includes many leaping, turning and balancing movements. A successful ballet performance involves movement continuity, rhythm, expressiveness, aesthetic positioning, dynamic versatility, and stage presence (Minden, 2005). Ballet dancers need strength, flexibility, balance abilities, and kinaesthetic awareness to accomplish such artistic movement (Minden, 2005). In the versatile practices of ballet, leaping is a highly skilled and intensive movement and often leads to the risk of injury. Ballet's simple ground échappé is quiet basic leaping practice. However, it is different from ordinary jumps such as counter movement jump. Ballet dancers have to land with hip turnout in order to achieve the classical dance posture. Such turnout at the hip joint could lead to so called "screwing the knees" which place considerable stress on the knee joint (Milan, 1994). Moreover, during repetitive leaping, the knee would sustain great stress. Over a long period of time, accumulated muscular fatigue may change their movement patterns. The anterior knee pain is a common syndrome among ballet dancers. Knee injuries account for 14-20% of all ballet injuries (Milan, 1994). The purpose of this study was to investigate the differential effect of physical condition on the impact force and the knee joint kinematics and kinetics of the dominant leg during ballet's simple ground échappé in ballet dancers with and without anterior knee pain. It was hypothesized that there would be difference found between dancers with and without anterior knee pain in different physical conditions.

METHOD: Seven female ballet dancers with anterior knee pain (age: 18.6 ± 0.5 years; height: 161.7 ± 2.9 cm; mass: 50.7 ± 4.2 kg) and seven ballet dancers without anterior knee pain (age: 18.1 ± 0.4 years; height: 158.4 ± 3.2 cm; mass: 50.9 ± 5.1 kg) voluntarily participated in this study. Subjects with or without anterior knee pain were estimated by a physical therapist according to Laprade (2002). Informed consent was obtained from all subjects. This study was approved by the Ethical Committee of the University. A warm-up of ten-minute stretching was performed prior to the testing protocol. Each subject performed ballet's simple ground échappé for twenty sets (Figure 1_a~e). The leaping pace was controlled by a metronome (75 times/minute). The rating of perceived exertion (RPE) was obtained after testing. The RPE of subjects with or without anterior knee pain were 14.3 ± 1.3 and 14.0 ± 1.0, respectively, in which the exertion level was between some hard and hard. Subjects' knee isometric contraction was also test on Biodex (Biodex Medical Systems, Inc. Shirley, NY, USA). The knee extension peak torque of subjects with and without anterior

knee pain was 3.2 ± 0.8 and 3.1 ± 0.9 Nm/kg. The knee flexion peak torque of subjects with and without anterior knee pain was 1.3 ± 0.2 and 1.4 ± 0.3 Nm/kg, respectively.
The movement data were collected with ten Eagle cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) at 200 Hz sampling rate which were positioned around the performance area. Reflective markers were placed in a modified Helen Hayes configuration to identify the segment motion. Cameras were synchronized to two force platforms (AMTI Inc., Watertown, MA, USA) which sampling rate was 2000 Hz. One platform collected the right leg data, and another collected the left leg data. Both kinematic and kinetic data were recorded in EVaRT software (Version 4.6, Motion Analysis Corporation, Santa Rosa, CA, USA).

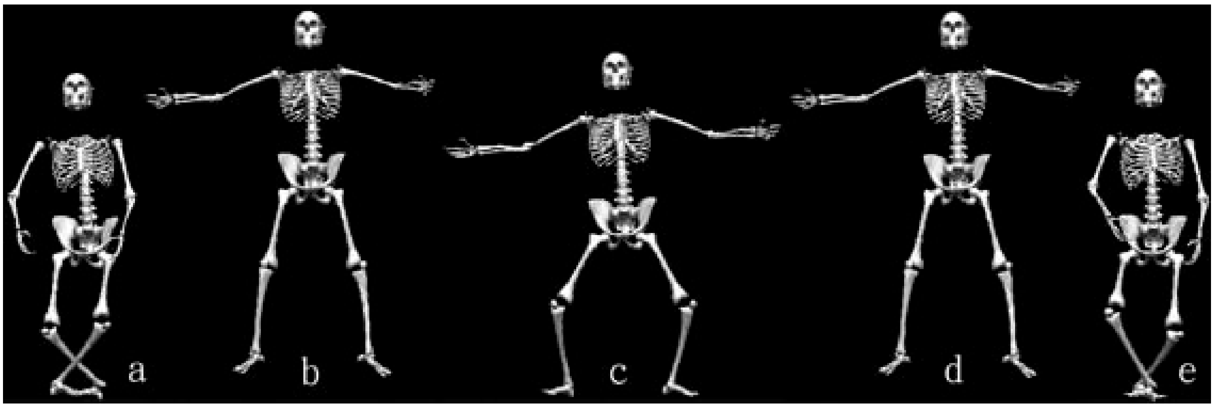

Figure 1. a~e: A set of the ballet's simple ground échappé.

The data were analyzed in The Monitor Monitor software (Version 8, Innovative Sports Training, Inc., Chicago, IL, USA). The dominant leg, determined in relation to the foot normally used to kick a ball, was analyzed for all subjects. In order to compare the easy and hard conditions of ballet's simple ground échappé, we defined the first and last three sets to refer to the easy and hard conditions, respectively. Subjects can only step on one force platform with one leg when splitting legs (Figure 1_c) in which contact phase was defined. During the contact phase, the knee joint kinetics was analyzed. The moment and power of knee joint was calculated from the inverse dynamics. The contact phase was then separated eccentric and concentric phases which determined using maximum knee flexion threshold. The contact phase, defined as the time from foot contact to take off from the ground, was analyzed. The instant of foot contact and take off during the contact phase was determined using a 30 N vertical ground reaction force (vGRF) threshold, respectively. Kinetic variables were normalized by each subject's body weight. Mix design two-way (2 groups × 2 conditions) ANOVAs was used to compare the differences between groups and conditions. The significance level was set at α=.05. The *post-hoc* analysis was performed with the Bonferroni test. The effect size was calculated on all variables.

RESULTS: All subjects had significantly greater peak net knee moment (P=.026) and peak power generation (P<.001) in easy condition than in hard condition. They took significantly longer time during the concentric phase (P=.003) and the total contact phase (P=.012) in hard condition than in easy condition (Table 1). There was no significant difference found between dancers with and without anterior knee pain in the rest of variables. The between-conditions effect sizes were shown in Table 2. Dancers without knee pain showed large effect size on eccentric time (effect size=1.08). The between-dancers effect sizes were shown in Table 3. The peak knee power absorption showed medium effect size between dancers with and without knee pain in both easy and hard condition (effect size=0.47 and 0.52, respectively). The peak knee moment showed large effect size between dancers with and without knee pain in hard condition (effect size=1.03). There were medium effect size on eccentric time in easy condition (effect size=0.46) and large effect size on concentric time in hard condition (effect size=0.86).

Table 1. Knee joint kinetics and kinematics of the dominant leg during first contact phase.

		Easy Condition	Hard Condition
Without Knee Pain	Peak vGRF (BW)	1.38 ± 0.13	1.39 ± 0.28
	Peak KneeMom (Nm/kg)	1.57 ± 0.21	1.39 ± 0.12 *
	Peak KneePowAbs (W/kg)	6.79 ± 2.07	7.10 ± 1.50
	Peak KneePowGen (W/kg)	6.49 ± 1.32	4.48 ± 0.99 *
	Peak KneeFlexAng (deg)	83.10 ± 7.75	82.16 ± 7.84
	KneeFlexAngCONT (deg)	16.41 ± 7.60	15.56 ± 6.72
	TimeEcc (ms)	235 ± 12	261 ± 32
	TimeCon (ms)	243 ± 11	303 ± 64 *
	TimeCONT (ms)	479 ± 14	565 ± 87 *
With Knee Pain	Peak vGRFMax (BW)	1.53 ± 0.24	1.47 ± 0.26
	Peak KneeMom (Nm/kg)	1.65 ± 0.26	1.56 ± 0.20 *
	Peak KneePowAbs (W/kg)	7.77 ± 2.07	7.91 ± 1.64
	Peak KneePowGen (W/kg)	6.61 ± 1.76	4.85 ±1.06 *
	Peak KneeFlexAng (deg)	83.23 ± 7.08	83.03 ± 4.66
	KneeFlexAngCONT (deg)	14.78 ± 3.83	16.17 ± 2.41
	TimeEcc (ms)	247 ± 35	268 ± 61
	TimeCon (ms)	239 ± 35	261 ± 25 *
	TimeCONT (ms)	488 ± 69	530 ± 78 *

* Significance found between easy and hard condition. KneeMom=Knee moment; KneePowAbs= Knee power absorption; KneePowGen= Knee power generation; KneeFlexAng=Knee flexion angle; KneeFlexAngCONT=Knee flexion angle at contact; TimeEcc=Time of eccentric phase; TimeCon= Time of concentric phase; TimeCONT=Time of total contact phase.

Table 2. Effect Size between easy and hard conditions

	Without Knee Pain	With Knee Pain
Peak vGRFMax (BW)	0.05	0.24
Peak KneeMom (Nm/kg)	1.05 *	0.39
Peak KneePowAbs (W/kg)	0.17	0.07
Peak KneePowGen (W/kg)	1.72 *	1.21 *
Peak KneeFlexAng (deg)	0.12	0.03
KneeFlexAngCONT (deg)	0.12	0.43
TimeEcc (ms)	1.08 *	0.42
TimeCon (ms)	1.31 *	0.72 *
TimeCONT (ms)	1.38 *	0.57 *

* Medium to large effect size.

Table 3. Effect size between dancers with and without knee pain.

	Easy Condition	Hard Condition
Peak vGRF (BW)	0.78 *	0.30
Peak KneeMom (Nm/kg)	0.34	1.03 *
Peak KneePowAbs (W/kg)	0.47 *	0.52 *
Peak KneePowGen (W/kg)	0.08	0.36
Peak KneeFlexAng (deg)	0.02	0.13
KneeFlexAngCONT (deg)	0.27	0.12
TimeEcc (ms)	0.46 *	0.14
TimeCon (ms)	0.15	0.86 *
TimeCONT (ms)	0.18	0.42

* Medium to large effect size.

DISCUSSION: The purpose of this study was to compare the differential physical condition during ballet's simple ground échappé in ballet dancers with and without anterior knee pain. Major findings indicated that ballet dancers exerted greater knee extensor moment and power generation in hard physical condition during simple ground échappé. The repetitive leaping made ballet dancers feel hard to keep up the rhythm and aesthetic positioning. Therefore, more knee extensors effort had been elicited to accomplish the movement.

Ballet dancers spent more time on ground contact and concentric movement in the hard physical condition. Maybe they were trying to generate more effort from ground reaction force and exertion powers from the knee extensor muscles for jumping off. On the other hand, ballet dancers also spent more time on eccentric movement during landing in hard physical condition. We suggested that they can get more power absorption from knee extensor muscles to decelerate the body during landing.
In the study, we did not find any statistical difference between ballet dancers with and without anterior knee pain. However, there were some rather great mean differences observed with medium to large effect size. Dancers with knee pain landed with greater power absorption from the knee joint than dancers without knee pain. Also, dancers with knee pain had greater knee extensor moment than dancer without knee pain in hard physical condition. Furthermore, they spent more time on eccentric movement than dancers without knee pain in easy physical condition. Combining the aforementioned evidence, we suggested that ballet dancers with knee pain tended to protect their knees with greater knee power absorption during simple ground échappé compared to dancers without knee pain.
In this study, no difference was found on knee joint positions. This meant that differential physical condition or knee pain did not affect the outcome performance on landing during simple ground échappé. Even dancers had knee pain or in hard physical condition; they can still maintain the adequate body positioning to achieve the requirement of aesthetic position in ballet dance. However, the changes on internal knee execution pattern had been found.
Further investigations into the muscular activities would guarantee the internal interpretation of the neuromuscular performance of knee joint muscles during simple ground échappé. Although only a few variables showed statistical difference in this study, some variables showed medium to large effect size. The insignificant findings may be due to the low subject number. Recruiting more subjects is guaranteed to further study.

CONCLUSION: This study explored the effect of physical condition in ballet dancers with and without anterior knee pain. Ballet dancers exerted greater knee extensor moment and power generation in hard physical condition during simple ground échappé. Ballet dancers with knee pain tended to protect their knees with greater knee power absorption during simple ground échappé compared to dancers without knee pain. Strengthening ballet dancers' knee extensors was advocated to be a part of training program.

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THE EFFECTS OF A CLOTH WRAP IN STABILIZATION OF THE ANKLE

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This study aimed to examine the effects of the cloth wrap in ankle stabilization, since there is limited research of cloth wrap ankle stabilization. Twenty subjects (13 female, 7 male) were utilized in this study. Three conditions were performed on each ankle (Baseline, Wrapped, and Unwrapped). Results showed the only significant difference ($p<0.01$) found were the main effects for condition (Baseline vs. Wrapped, vs. Unwrapped). A significant interaction of condition by gender was also found ($p=0.034$). None of the other interactions were significant ($p>0.05$). It was concluded that the cloth wrap may have decreased the subject's proprioception. Females also stabilized out faster than males, possibly due to a lower center of gravity (COG).

KEYWORDS: stability, force platform, ankle sprain, center of gravity (COG).

INTRODUCTION: Ankle sprains are one of the most common injuries in athletics, representing from 38-50% of total sport injuries (Abián-Vicén 2008**)**. Despite the high prevalence of ankle injuries, controversy still exists on the best way to prevent this common injury (Mickel et al. 2006). The most common argument exists in whether to utilize prophylactic taping techniques or bracing. Both have the benefits of preventing plantar flexion and inversion, the common mechanism of injury in ankle sprains (Mickel et al 2006). However, taping especially has its downfalls. Taping procedures, regardless of the technique, have been known for loosening with physical activity (Mickel et al 2006, Nigg et al 2000). The cost-effectiveness of tape has also been scrutinized (Abian-Vicen 2008, Childs 2007).

The research on ankle bracing and taping techniques and their effects on preventing ankle sprains is plentiful; however, little research has been conducted on the effectiveness of a cloth ankle wrap in stabilizing the ankle. Because balance is such an important factor in all sports, whether in dynamic or static movement, this study utilized balance testing to determine the effectiveness of a cloth wrap in stabilizing the ankle joint. It is important to note that with dynamic balance comes the increased risk of injury. Upper body posture and stability directly affect the lower body due to the kinetic chain. Furthermore, to protect internal structures such as ligaments and cartilage from injury, a lower limb needs to be stabilized when loaded (Ackland et al 2009). Failure of muscles or ligaments to control joint posture results in excessive loading causing considerable motion between the two articulating bones of the joint, consequently placing great strain on the stabilizing structures (Ackland et al 2009). The purpose of this experiment was to examine the effect a cloth ankle wrap has on ankle stabilization while performing a static balance test.

METHOD: Twenty college students were recruited for this study (age: 21.5 ± 1.5 years), 13 female, 7 male. Ten subjects had previously sustained ankle injury and ten subjects had not. Subjects were asked to fill out a survey and a PAR-Q form before participating in the study. Written informed consent was provided by all twenty participants for the study which was approved by the University Institutional Review Board (Human Subjects Proposal Number: HS09-315).

Subjects' weight and height were measured and recorded (height: 169.3 cm ± 9.2 cm; weight: 735.4 N ± 138.8 N). The subjects took off their shoes and the balance for both ankles was analyzed using a force platform (AMTI Model OR6-5; Watertown, MA). Three conditions were performed for each ankle: Baseline (B), Wrapped (W), and Unwrapped (U), with B condition performed first. From then on, the order of the conditions and sequence of the right or left ankle were randomized with a coin flip. The force platform was zeroed in the loaded state and the subjects placed the determined foot in the center of the platform and balanced for twenty seconds. The W and U conditions were performed while standing on a dyna-disc (Exertools, Rohnert Park, CA) to create perturbations (Figure 1). The W trials utilized the cloth ankle wrap for stabilization. The subjects sat on a table, extending their leg and positioning their foot at a 90° angle. The cloth was wrapped around the ankle using one heel lock and one figure eight (see Figure 2). For additional support, two heel locks and two figure eights were applied with adhesive tape over the cloth wrap (Prentice 2006). The U conditions did not utilize the cloth ankle wrap for stabilization. The subjects were allowed freedom to move their arms and opposite leg positioning as they chose. The right hip and knee started in approximately 5° of flexion and the ankle was in the neutral position. However, these positions fluctuated as necessary as the subjects tried to maintain their balance on the dyna-disc. All data were recorded using the AMTI NetForce software (version 2.1.1, Watertown, MA). Standard deviations of the subjects' movement as assessed by the vertical ground reaction forces were compared amongst each other as the dependent variable. Standard deviations were determined by comparing the movement of a subject over a period of twenty seconds. The standard deviations of these data was used as the means of comparision among conditions.Statistical analyses of the data were carried out in SPSS 17.0 (Chicago, IL). A Mixed ANOVA with Repeated Measures on condition (B, W, U) with non repeated measures on Gender, Injury Presence/Absence, and Side. The criterion for significance was set at an alpha level of $p \leq 0.05$.

Figure 1. Subject balancing on the dyna-disc.

Figure 2. Application of the cloth wrap.

RESULTS: Analysis of variance found no differences across the injuries, left and right sides, or genders. The only significant difference ($p < 0.001$) found were the main effects for condition (B vs. W vs. U). There were no significant differences across the conditions, genders, or right versus left foot ($p > 0.05$). In addition, a significant interaction of condition by gender was found ($p = 0.034$). None of the other interactions were significant ($p > 0.05$). The interaction of gender and condition and their values are illustrated in Figure 3.

DISCUSSION: The similarities during baseline conditions between males and females may be attributed to response time. Because there were no outside perturbation factors, the subjects were able to gain postural control and stability relatively quickly within the twenty-second time duration. However, when the dyna-disc was added to create perturbations in the U and W conditions, the differences in fluctuation were apparent between the male and female subjects. As shown in Figure 3, the female subjects had lower standard deviations over the U and W conditions (3.3667 and 3.8944 respectively) than the male subjects (4.7105 and 5.5744 respectively). A possible explanation might be the lower location of the center of gravity (COG) for women compared to men (Prentice, 2004). The lower height of their COG may give women an advantage over men in balancing sports (Alexander 1997). With the lower COG, the female subjects may have been able to gain control over their movement quicker than their male counterparts.

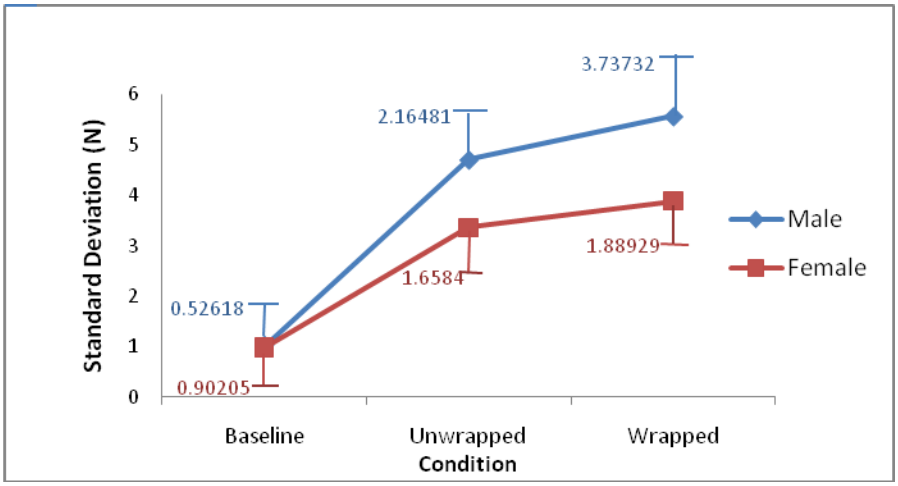

Figure 3. The standard deviation of male and female single-leg balancing over a twenty-second duration for conditions of Baseline, Wrapped, and Unwrapped.

Based upon the results of the main effects for condition (B vs. W vs. U), it is clear to see the W condition had higher standard deviation values than the U condition. Bracing and wraps have been shown to decrease proprioception (Nigg et all 2000), which may be why the W condition's standard deviations were higher for both subjects. The wrap only stabilizes the ankle joint, and because of the kinetic chain the subject's movement was more sporadic in the W condition as compared to the U condition.

The low subject number could the reasoning behind the non-significant findings for the right and left side, the presence or absence of injury, and the differences within genders. This study should be expanded to fully analyze these differences, in addition to gaining knowlege on the different stability maintenance techniques between male and female subjects.

CONCLUSION: There is still controversy existing about what method is best for prophylactic prevention of ankle sprains. This study was undertaken to observe the effectiveness of a cloth ankle wrap in stabilizing the ankle during a balance test. It was determined that the cloth ankle wrap was effective in stabilizing the ankle, in turn helping the subjects maintain their balance. In addition, results showed that women were able to maintain stability better in subsequent trials.

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HIP ROTATION RANGE OF MOTION AND ITS IMPACT ON LOWER LIMB ALIGNMENT ON LANDING

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The purpose of the present study was to compare lower limb alignment at initial ground contact between groups with normal and abnormal hip rotation range of motion. Male (n=8) and female (n=8) subjects performed an maximal drop jump diagonal side cut task ten to the left and ten to the right. Lower limb alignment was assessed through knee angle, hip angle, ankle angle, thigh rotation and shank rotation at initial foot contact. One significant difference was reported between groups for the knee angle variable on the non dominant side. This indicates that the only the knee angle variable is affected by unbalanced hip rotation range of motion and on the non-dominant side.

KEY WORDS: Anterior Cruciate Ligament, joint angles, limb dominance, segment rotations

INTRODUCTION: Anterior cruciate ligament (ACL) injuries are well recognised as one of the most common and serious sports injuries with upwards of 250,000 ACL injuries in the United States each year (Boden *et al.* 2000). Seventy two percent of non-contact ACL injuries occur at or shortly after foot strike (Myklebust *et al.* 1997, Olsen *et al.* 2004). Lower limb alignment at the moment of foot strike has been shown to effect valgus loading at the knee (McLean *et al.*, 2004a; Zeller *et al.*, 2003) with greater valgus loads increasing ACL injury risk. Numerous factors can alter lower limb alignment at landing such as hip joint neuromuscular control, strength and range of motion (ROM). Neuromuscular control at the hip has been suggested to be functional as a means of countering ACL injury inducing valgus loads (Besier *et al.* 2003, Lloyd and Buchanan 2001, Zhang and Wang 2001). Decreased ROM at the hip has also been linked to injury risk for conditions such as lower back pain (Verrall *et al.* 2007, Ireland and Wall 1990). Hip rotation ROM has also been shown to differentiate ACL injured subjects from the general population (Gomes et al 2008); it is unclear if this would still be the case prospectively. Lower limb alignment at landing and hip function have been previously investigated in terms of their affect on valgus loading and ACL injury risk however there is no indication of any interaction between the two. This study aimed to investigate differences in initial contact lower limb alignment dependent variables (knee angle, hip angle, ankle angle, thigh rotation, shank rotation) between two subject groups divided according to hip rotation ROM balance (independent variable) normal with external rotation <10° different from internal rotation (balanced) and abnormal with >10° difference (unbalanced). It was hypothesised that the group with abnormal hip rotation would display different landing alignment to that of the group with normal hip rotation ROM.

METHOD: Subjects included eight males (age 21 ± 3 yrs; height 1.79 ± 0.06 m; mass 76 ± 6 kg) and eight females (age 21 ± 2 yrs; height 1.68 ± 0.07 m; mass 64 ± 7 kg). Passive hip rotation ROM assessment took place prone on a plinth style bed (Harris-Hayes *et al.* 2007) using manual pelvic stabilisation and a bio-med gravity based inclinometer for measurement of shank orientation (Cibulka *et al.* 1998). Following ROM and maximum drop jump height assessment subjects completed 20 trials of a dynamic task. This involved dropping from a 0.30 m bench, and performing an immediate drop jump up to reach and touch a target which was suspended at their maximum drop jump height. The suspended target triggered a directional cueing system which indicated which direction the subject had to run diagonally to on landing (10 left 10 right), set up is shown in Figure 1.

Data Collection: A six-camera high-speed motion analysis system (Eagle; Motion Analysis Corp., Santa Rosa, CA) (500Hz) was synchronised with an AMTI dual force platform system (1000 Hz). 43 retro-reflective markers were secured on the asis, psis, sacrum, iliac crest, greater trochanter, medial and lateral epicondoyle and malleolus, upper and lower calcaneous, 2nd and 5th metatarsal of both legs four marker clusters were also placed on the thigh and shank. Each subject stood for a static trial prior to full data collection.

Data Analysis: Hip rotation groupings were defined as mentioned according to hip rotation ROM. The abnormal grouping was unidirectional demonstrating an increased level of external rotation. These groups had significantly different internal external rotation ratio's with

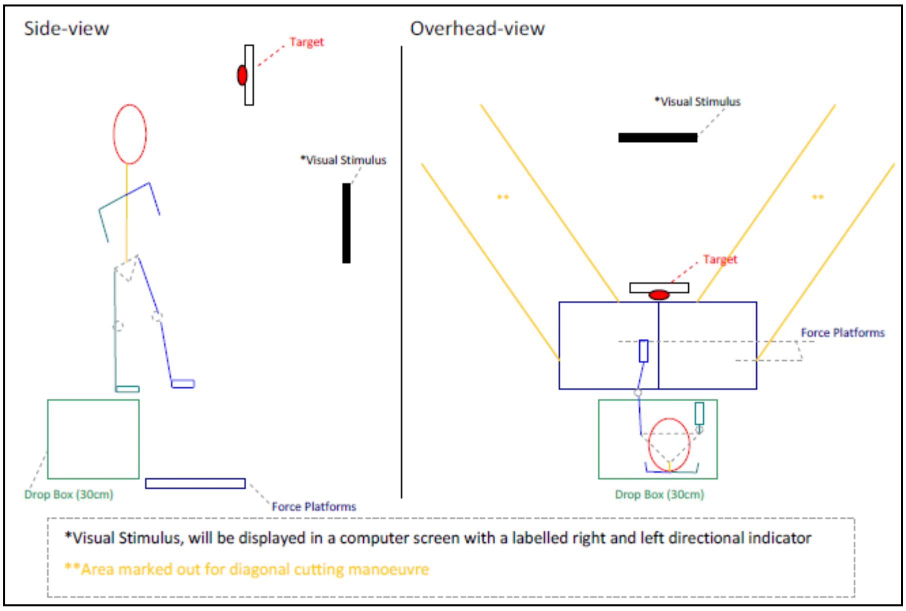

Figure 1. Dynamic Task Set Up

a mean difference of 29°*. Knee hip and ankle angles were presented as anatomical angles thigh and shank rotations were presented as the difference between standing postition (static trial) and the position at ground contact. The differences between the hip rotation groups were assessed by a one way anova and Cohen's d was used as a measure of effect size. Pearson's correlation was also used to assess correlations between ROM internal external rations and the dependant variables.

RESULTS: Average values for all lower limb alignment variables recorded at ground contact are presented in Table 1 for both groups. One significant difference was found between the groups for the knee angle variable on the dominant leg. The group with abnormal hip rotation ROM demonstrated significatly more knee extension on landing in their non-dominant leg that the normal hip rotation ROM group. There was a very strong effect size also shown for this variable.

Table 1 Lower limb alignment variables at initial contact for both groups

	Normal Hip ROM		>Ext Hip ROM		*p*-value		Cohen's *d*	
(°)	Dominant	Non-Dominant	Dominant	Non-Dominant	Dominant	Non-Dominant	Dominant	Non-Dominant
Knee Angle	159	**156***	162	**168***	0.258 [a]	**0.003*** [f]	0.59	**1.78**
Hip Angle	131	122	126	122	0.181 [b]	0.871 [g]	0.73	0.08
Ankle Angle	133	133	125	125	0.309 [c]	0.305 [h]	0.55	0.55
Thigh Rotation	32	26	34	33	0.8 [d]	0.317 [i]	0.13	0.52
Shank Rotation	19	14	10	23	0.26 [e]	0.144 [j]	0.59	0.78

Power Analysis [a]= 0.196 [b]=0.259 [c]=0.166 [d]=0.057 [e]=0.195 [f]=0.897 [g]=0.053 [h]=0.168 [i]=0.162 [j]=0.302

When a Pearson's correlation was implemented on hip rotation range of motion and each dependent variable one distinct relationship was shown for knee angle on the non dominant leg with a correlation coefficient of 0.54*, significant at 0.030 (Figure 2).

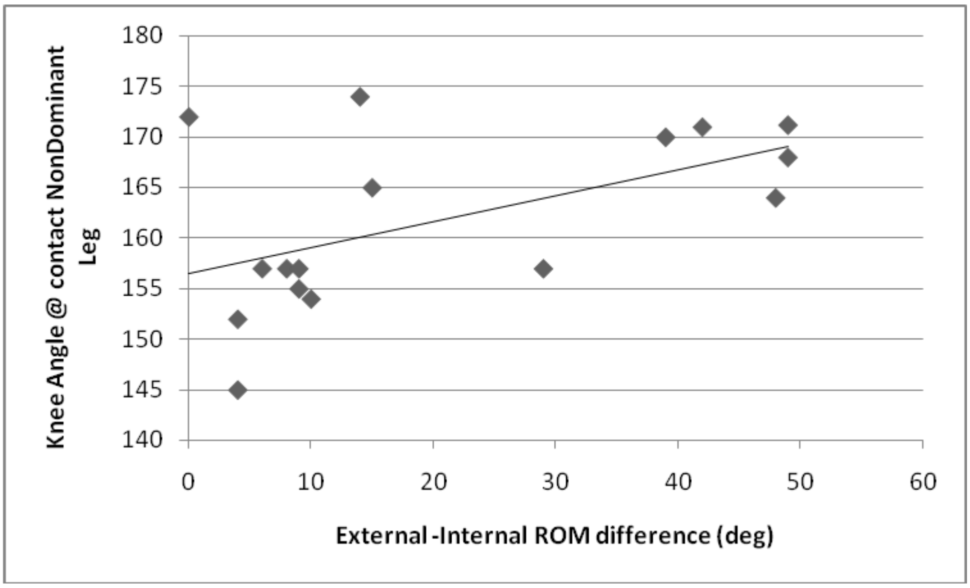

Figure 2. Knee Angle - Hip Rotation ROM Correlation

DISCUSSION: The purpose of this study was to investigate any differences in initial contact lower limb alignment variables (knee angle, hip angle, ankle angle, thigh rotation, shank rotation) between two subject groups divided according to hip rotation ROM. There was one significant difference shown between the two groups (non dominant knee angle) but overall the proposed hypothesis was not supported. There has been no previous research assessing the interaction between initial contact lower limb alignment and levels of hip rotation ROM. This investigation is therefore important in establishing that hip ROM does not have a significant effect on all of the landing alignment variables. The one variable that did show an interaction with hip rotation ROM was knee angle on the non dominant leg. Non-dominant knee angle at contact was significantly different between groups with a strong effect size and 29% of its variability was explained by each subject's hip rotation ROM (Pearson's correlation coefficient 0.54). The knee at ground contact in the abnormal hip rotation ROM group was significantly more extended which has been cited as a potential risk factor for ACL injury (Boden *et al.* 2000). The fact that this is only the case in the non-dominant limb is also interesting as limb dominance has also been assessed as a potential ACL injury risk factor It is thought that more ACL injuries occur on the non-dominant side but this has not yet been proven conclusively (Matava *et al.* 2002). Overall the proposed hypothesis was not supported as only one of five variables differed between groups. As stated in the introduction many factors affect lower limb alignment at landing; it is plausible that factors such as neuromuscular control and strength around the hip joint may demonstrate a stronger effect than the variables utilised in this study on lower limb alignment at landing. Additional measures that were not employed in this study which would have provided further depth to the investigation was valgus angle and joint moments, future analysis of those variables for the given data is planned. Areas that may alter lower limb alignment on landing and may merit future research include neuromuscular control, and strength at the hip. Variables such as muscle strength can be easily targeted in injury prevention interventions for the production of safer lower limb alignments at landing.

CONCLUSION: One variable from five lower limb alignment variables was shown to be significantly different when compared in terms of hip rotation, this variable was knee angle on the dominant leg. This relationship between knee angle and hip rotation is important as the more extended position adopted by those with abnormal hip rotation ROM may place them at an increase risk of ACL injury. Hip rotation ROM is easily targeted by stretching interventions which may act to decrease knee extension at ground contact and decrease injury risk. Future research is necessary to investigate the relationship between other hip joint functional measures such as strength and balance etc. This would assist in the development of other appropriate injury prevention programmes for the alteration of lower limb alignment at landing to decrease injury risk.

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HAMSTRING MUSCLE ACTIVATION DIFFERENCES BETWEEN GENDERS WHILE PERFORMING SINGLE LEG LANDINGS

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Women are 2-10 times more likely to incur an ACL injury compared to males. Hamstring neuromuscular differences between genders may contribute to genu valgum, a posture that is unfavorable to ACL integrity. The intention of this study was to examine hamstring activation patterns between genders when executing single leg landings. Ten male and ten female recreationally active subjects performed three repetitions of a single leg drop landing onto each leg. Surface EMG data were obtained from the medial and lateral hamstrings. A 2X2X2 repeated measures ANOVA indicated there were no main effects for gender, side, or muscle on the dependent variables ($p > 0.05$). Results of this study suggest that males and females exhibit similar medial/lateral hamstring neuromuscular activation strategies when landing from a jump.

KEYWORDS: anterior cruciate ligament, muscle activation, semi-membranosis, semi-tendinosis.

INTRODUCTION: Females incur a disproportionate rate of non-contact type ACL injuries when compared to their male counterparts (Hewett et al., 2005). Lower extremity neuromuscular mechanisms may be responsible for these injuries. Female athletes have been shown to land with greater dynamic knee valgus during sporting activities. Lower extremity muscle activity during dynamic movements may lead to posturing detrimental to ACL integrity (Myer et al., 2005). The purpose of this study was to examine the medial/lateral activation patterns of the hamstring musculature between genders when performing single leg landings from an elevated position.

METHODS: Ten male and ten female recreationally active college students were recruited from the campus of Northern Michigan University for this study. Subjects were excluded from the study if they reported less than at least 60 minutes of physical activity per week or previous history of lower extremity injury or disorder. Female mean age, height, and weights were; 22.5 ± 4.7 years, 168.75 ± 5.8 cm, and 65.60 ± 8.9 kg respectively. Male mean age, height, and weights were; 24.2 ± 4.0 years, 179.50 ± 10.5 cm, and 77.34 ± 15.0 kg respectively. The use of human subjects was approved by Northern Michigan University's Human Subjects Research Review Committee (# HS08-233). Informed consent forms and a joint pain questionnaire were reviewed and signed by each subject prior to data collection.

This study used bipolar surface electrodes connected to an amplifier (MP 150, BioPac Systems Inc, Goleta, CA) to determine muscle activity when landing from an elevated position. Data were collected at a sample rate of 1000 Hz or a period of 5 seconds. A 10-500 Hz band pass filter was employed. Data were then saved to a personal computer (IBM Thinkpad) for later analysis. Prior to interpretation, the raw electromyography (EMG) data were rectified using the root mean square. The rectified data was then integrated by averaging over every 20 samples. Analysis of this data was performed using AcqKnowledge 3.9.1 (BioPac Systems Inc, Goleta, CA, USA).

The electrode sites were prepared by shaving leg hair with a disposable razor, abrading the epidermal skin layer, and swabbing the sites with isopropyl alcohol to reduce impedance of the skin to <5-kilo ohms (kΩ). Disposable self-adhesive Ag/AgCl snap electrodes (Noraxon, Scottsdale, AZ, USA) were secured, bilaterally, over the muscle bellies of the biceps femoris and semi-tendinosis. The conductive surface of each electrode measured 1 cm, with an inter-electrode distance of 2 centimeters (cm). Ground electrodes were placed bilaterally, over the ipsilateral medial and lateral tibial condyle.

Preceding the landing trials, normalization of EMG data were performed relative to a maximal isometric contraction (MVC) of the hamstring musculature for comparison between the subjects and side (left/right). Subjects were asked to sit on an Isokinetic Dynamometer (Biodex Shirley, NY, USA) with their knees and hips flexed to 90 degrees. EMG data were captured while the participant was encouraged to perform a single leg maximal hamstring isometric contraction for 5 seconds on each lower extremity. Following the MVC's, each subject was allowed to rest while standing for approximately 5 minutes.

A single force plate (OR6-7-2000, AMTI, Watertown, MA, USA) was used in this study to determine the moment of initial contact when landing occurred. Vertical ground reaction forces (VGRF) during the landings were recorded at a sampling rate of 1000 Hz utilizing Netforce 2.0 software (AMTI Watertown, MA, USA). Both EMG and VGRF signals were chronologically synchronized to one signal during data collection for later analysis. Time to peak integrated electromyography (IEMG) for the four muscles was calculated as the time when muscle activity reached peak amplitude following initial contact.

Drop landing trials required the subjects to hang from an elevated position by both hands. A 50 cm multipurpose straight bar was suspended from an adjustable zinc coated chain that was secured to the building structure located directly above the subject. For each subject, the bar was adjusted so that the subject's feet were 33 cm above the force platform while hanging, with the plantar surface of the feet parallel with the force platform. A total of six trials of the landing task were performed (3 onto each leg). When instructed, the subject would release their hands from the straight bar and drop onto the force platform. Landing side (right vs. left) was randomly assigned for each trial. Subjects were asked to maintain balance upon landing to the best of their ability.

The dependent variables observed in this study were time to peak integrated EMG amplitude from initial landing contact, mean integrated EMG amplitudes, and percent MVC for the medial and lateral hamstring (semi-tendinosis and biceps femoris). Means and standard deviations were calculated for the dependent variables. A 2×2×2 (gender × side × muscle) mixed design ANOVA, where the landing side and muscles were repeated measures, was used to evaluate the main effects of each independent variable on biceps femoris muscle activation as well as the main effects on semi-tendinosis muscle activation.

RESULTS: Results revealed no significant differences between men and women for mean IEMG activity following initial contact ($p = 0.412$). No significant differences were found when examining mean IEMG activity after initial contact between side ($p = 0.722$). Mean IEMG activity subsequent to initial contact showed no significant differences between the semi-tendinosis and biceps femoris ($p = 0.252$). A summary of IEMG values are presented in Table 1.

Peak IEMG following initial contact was expressed as a percentage of the subject's MVC. Semi-tendinosis and biceps femoris percent MVC were not shown to be significantly different between males and females ($p = 0.879$). No significant differences were found when examining percent MVC at peak IEMG following initial contact between landing side ($p = 0.959$). The two hamstring muscles examined in this study were shown to not be statistically different when comparing percent MVC of peak IEMG after initial contact ($p = 0.181$). Table 2 shows peak IEMG values.

Time to peak IEMG after initial contact examination between genders confirmed that there was no statistical difference ($p = 0.619$). There were no differences between landing leg when analyzing time to peak IEMG after initial contact ($p = 0.986$). Both the semi-tendinosis and biceps femoris demonstrated peak IEMG activation below the level of significance ($p = 0.696$). Actual time to peak IEMG values are reported in Table 3.

Table 1. Mean IEMG activity shown in volts.

	Mean IEMG Muscle Activation			
	Left Biceps Femoris	Right Biceps Femoris	Left Semi-Tendinosis	Right Semi-Tendinosis
Male	0.01375 ± 0.00867	0.01031 ± 0.00406	0.01404 ± 0.00838	0.01515 ± 0.01045
Female	0.00968 ± 0.00470	0.01002 ± 0.00378	0.01214 ± 0.00441	0.01331 ± 0.00544

Table 2. Peak IEMG shown as % MVC following initial contact.

	Peak IEMG Muscle Activation (% MVC)			
	Left Biceps Femoris	Right Biceps Femoris	Left Semi-Tendinosis	Right Semi-Tendinosis
Male	50.8 ± 33.5	43.0 ± 22.2	49.4 ± 17.5	42.8 ± 20.1
Female	40.5 ± 21.4	47.2 ± 23.0	45.1 ± 13.4	50.3 ± 17.9

Table 3. Time (seconds) from initial contact when landing a jump until peak muscle activity.

	Time-to-Peak Muscle Activation			
	Left Biceps Femoris	Right Biceps Femoris	Left Semi-Tendinosis	Right Semi-Tendinosis
Male	0.4219 ± 0.1357	0.3670 ± 0.1783	0.3653 ± 0.1598	0.3040 ± 0.1360
Female	0.3222 ± 0.2097	0.3591 ± 0.1550	0.2996 ± 0.1907	0.3930 ± 0.1486

DISCUSSION: Disproportionate medial/lateral quadriceps muscle activation differences between genders have been identified, and could potentially explain the ACL injury rate disparity between the genders (Myer et al., 2005). In addition, Subsequent to landing, unbalanced medial/lateral hamstring activation may further jeopardize knee joint positioning. Activation of the lateral hamstrings prior to, or with greater force than that of the medial hamstrings could load the lateral tibio-femoral joint and leave the medial joint line open, leading to genu-valgum, and vulnerable to ACL injury. The combination of medial/lateral muscle activation disparities in both the quadriceps and hamstrings could increase this medial joint line space. The findings of this investigation do not support our hypothesis, suggesting another mechanism may be responsible for the disproportionate ACL injury rate identified between the genders.

Participants from the current study also performed bilateral lower extremity landings to be used for a separate investigation by Abe and coworkers (2009) comparing 3-dimensional (3D) kinematics during single leg and bilateral leg drop landings between genders. While in conflict with the kinematic findings of Chappell and others (2005) females demonstrated greater peak

knee flexion angles during the bilateral landing procedure when compared to their male counterparts, despite a lack of significance. These results suggest that females utilize different landing patterns to attenuate vertical loads when landing with both lower extremities, while both genders may use similar landing postures when landing on a single limb. The comparable knee angles seen with single leg landings in Abe's study may have contributed to the lack of differences in hamstring muscle activity results between genders seen during the current study. The greater knee angle differences noticed between genders during the bilateral leg landings may have equally resulted in different hamstring activity. Further examination is warranted to determine if hamstring activity differs between males and females during bilateral drop landings.
A power analysis was performed following the completion of this study to determine if a type II error existed. A beta value of 0.45 was found; indicating that a type II error was possible (Faul et al., 2007). Future studies should utilize a greater number of subjects to minimize the chances of committing a type II error, while attention is directed at examining 3D knee kinematics in conjunction with lower extremity muscle EMG data to establish if females utilize different landing techniques than males, and how these techniques are influenced by muscle activity.
The similar hamstring activity found between genders in this study may impact the course of ACL injury prevention strategies. Future research should be conducted to confirm if hamstring activity is similar during bilateral leg landings. This may help rule-out medial/lateral hamstring activation differences as a contributor to the ACL injury rate disparity between genders. Instead, attention should be aimed toward other neuromuscular mechanisms, where gender differences were shown to exist, contributing to dynamic knee valgus (Zazulak et al., 2005).

CONCLUSION: The results of this study suggest that for the muscles studied, males and females possess similar neuromuscular activation strategies when performing single leg landings. These muscles therefore would not produce a gapping of the medial joint line about the knee. It appears that variations in medial/lateral hamstring muscle activation are not a contributor to the ACL injury rate disparity between gender when considering single leg landings. The attention of injury prevention professionals should therefore be directed at other causes of dynamic genu-valgum in recreational active females.

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THE INFLUENCE OF TWO DIFFERENT BRACES ON LATERAL PATELLAR DISPLACEMENT – A CADAVERIC STUDY

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KEYWORDS: patellar bracing, patellar kinematics, patellofemoral joint, patellar tracking

INTRODUCTION: Patellofemoral pain syndrome (PFPS) often occurs in young and physically active athletes (Taunton et al., 2002, Adirim & Cheng, 2003). It is generally accepted that a cause of PFPS is a malalignment of the patellofemoral joint. Bracing supply is commonly used for the treatment of PFPS. Several studies have shown that patellar bracing and taping improved PFPS (Lun et al., 2005, Warden et al., 2008). Crossley et al. (2009) found a reduced lateral patellar displacement and a decreased mean pain (mean pain was recorded on a 100 mm visual analog scale during single-leg squats) after patellar taping. A more medial displacement of the patella and a decrease in patellofemoral stress could lead to less patellofemoral pain (Powers et al., 2004). A more medial patellar displacement could result in a more centered patella. Therefore, the purpose of this study was to investigate the effect of two different braces on the alignment of the patella.

METHOD: Six fresh frozen cadaveric legs (3 subjects, age 66-72 years) were thawed for 24 h at room temperature. Apart from separating the femur head from the shaft, which was armed with a fixture, the legs were not dissected. The lower leg was fixed with belts and held in a vertical position. With the aid of the fixture each leg underwent 10 flexion-extension cycles through a range of 45° to 0°. During a flexion-extension cycle the thigh muscles were strained using a tighten strap which was armed with nails (inserted in the muscles) and fixed on the jig. The legs were tested in a non-braced condition followed by two conditions with braces. Two different braces were chosen: Patella Pro (PP) (Otto Bock GmbH, Germany) and a common elastic brace (BA) (Genutrain P3 Bauerfeind AG, Germany). Kinematic data were obtained by using Vicon Nexus with 5 Cameras (100 Hz) (Version 1.4.115, Vicon Motion Systems Limited, United Kingdom). Bone pins were screwed into tibia, femur and patella to minimize the influence of skin movements. To avoid a skin-pin impingement of the patellar bone pin the skin was incised along the line of motion. Each bone pin was armed with an array of three retroreflective markers. To define anatomical reference systems anatomical landmarks were pointed using a bar of 20 cm length attached with three retroreflective markers and related to their segmental bone pin. A mathematical model was built in Vicon Bodybuilder (Version 3.6, Vicon Motion Systems Limited, United Kingdom) to calculate the lateral patellar displacement (LPD) in relation to the femurs medio-lateral axis. The LPD was quantified during extension of the knee joint relative to the non-braced condition. The mean displacement of 10 extensions was obtained close to 0° for each condition. To find differences in LPD between BA and PP conditions we used a non parametric Wilcoxon signed-rank test for repeated measurements with a significance level of $P < 0.05$.

RESULTS: In the non-braced condition the position of the patella was lateral in relation to the center of the medio-lateral axis of the femur over the full flexion-extension cycle. The patella was more medial close to the maximal knee extension. The results for the LPD in relation to the non-braced condition are shown in Figure 1. For the PP brace the average LPD close to maximal knee extension was more medial by 0.86 mm ± 0.90. For the BA condition the average LPD close to maximal knee extension was more lateral by -0.73 mm ± 1.41. Therewas a significant difference in lateral displacement between PA and BA ($P = 0.028$).

DISCUSSION: The purpose of the study was to investigate the influence of patellar bracing on the lateral patellar displacement. The motion in the knee joint is a coupled movement between the tibiofemoral joint and the patellofemoral joint (Li, 2007). Compared to the literature (Koh et al., 1992; Brossmann et al., 1993; Varadarajan et al., 2010) the lateral displacements are similar to those reported in our study. While the use of cadavers is limited due to the lack of physiological muscle contractions the result indicates a life-like patellar motion in comparison to the literature. Small differences can be explained by relating the motion to different axis systems and by the complexity of the patellofemoral joint. The aim of a brace is to center the patella in the trochlea groove. Powers et al. (1999) found no significant differences in LPD between a non-braced condition and Bauerfeind Genutrain P3 brace. Our results showed even a slight lateral LPD in the BA condition. In the PP condition the LPD was more medial. These findings are supported by Crossley et al. (2009) with similar results to our study. They reported a significant reduced lateral displacement and decreased pain after taping the patella. In agreement to this Lun et al. (2005) demonstrated less pain during wearing a patella brace.

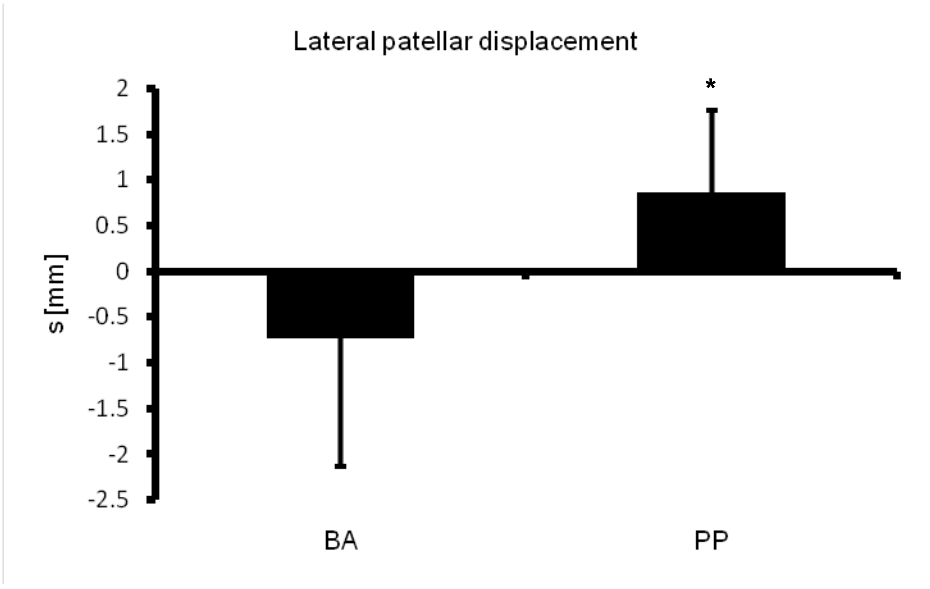

Figure 1. Positive values indicate a more medial displacement; negative values indicate a more lateral displacement in relation to the non-braced condition. The position of the patella in the non-braced condition is lateral in relation to femurs medio-lateral axis. *There is a significant difference between PP and BA (P = 0.028).

CONCLUSION: Our study showed the influence of bracing on lateral patellar displacement (LPD). We found a more medial LPD after bracing with Patella Pro (PP) (Otto Bock GmbH, Germany) and in contrast to this a more lateral LPD after bracing with the Genutrain P3 (BA) (Bauerfeind AG, Germany). Compared to studies which investigated the effect of bracing on patellofemoral pain syndrome (PFPS) our findings suggest that the use of the BA brace might not be effective in reducing PFPS and the design of the PP brace provides prerequisites to reduce PFPS. To clarify the mechanism of the PFPS and to get more insight on the influence of bracing on the PFPS in sports further studies are required.

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EFFECT OF ACTIVE VS. PASSIVE END-RANGE DETERMINATION ON SHOULDER AXIAL ROTATION IN THROWER ATHLETES

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The effect of active or passive end-range determination on shoulder axial rotation is unclear on overhead-throwing athletes. Twenty-two healthy males were equally divided into athletes and non-athletes groups and their throwing arm was tested during internal and external arm rotation and on active and passive end-range determination conditions. The humeral and scapular 3D position were recorded at the shoulder rotational end-range and compared across groups using two-way repeated-measures ANOVA. No differences were found between groups for all humeral and scapular variables. The active internal Thoracohumeral (TH) and Glenohumeral (GH) arches were significantly (p=0.00) higher than internal passive TH and GH. At the end-range of external rotation athletes showed a scapula less in protraction (p=0.027) and less in scapular posterior tilt (p=0.00). External passive TH and GH were significantly higher than external active TH and GH.

KEYWORDS: humeral axial rotation; end-range determination; throwing shoulder

INTRODUCTION: Overhead-throwing athletes include throwers (e.g. baseball pitchers), swimmers, water-polo, handball and volleyball players. From a functional standpoint these sports produce repetitive overhead motions, that are discontinuous and ballistic in nature, and where the throwing arm is forcefully moved forward from maximal external rotation to near maximal internal rotation, while is kept in an elevation position. This mechanical demand seems to be in the origin of the adaptive changes described on rotational range-of-motion (ROM) pattern in the throwing shoulder. This pattern favours the increased external rotation (external rotation gain) and limited internal rotation (glenohumeral internal rotation deficit), while the range of the total arc of motion (external arc plus internal arc) remains unchanged (Myers, Laudner, Pasquale, Bradley, & Lephart, 2006). Altered shoulder mobility is thought to develop secondary to adaptive structural (bones, capsule and ligaments) changes to the glenohumeral joint.

Clinical shoulder assessment often includes ROM measurement of internal and external humeral rotation recorded via goniometry by placing the patient supine or in a sitting position with the arm at 90º of abduction (Myers, et al., 2006; Yamamoto, et al., 2006). In a supine position, the arm is rotated to the internal and external end-range while kept fully supported on a table. A posterior force applied by the examiner on the coracoid process and clavicle limit scapular motion, and the arm movement is assumed restricted to the glenohumeral joint. In a sitting position, patient holds or supports his/her elbow at a side while the arm is rotating around the long axis of the humerus (Boon & Smith, 2000; Ellenbecker, Roetert, Piorkowski, & Schulz, 1996). On both ROM testing conditions, the joint end-range is determined by the examiner according with the capsular end-feel (Awan, Smith, & Boon, 2002; Barlow, Benjamin, Birt, & Hughes, 2002; Reagan, et al., 2002) scapular liftoff (Warner, Micheli, Arslanian, Kennedy, & Kennedy, 1990) or the presence of pain (Andrews & Bohannon, 1989). Some studies suggest the use of an active self end-range determination on shoulder thrower assessment in order to collect information close to the specific patterns of external and internal rotation, during the arm throwing cycle (Ellenbecker & Roetert, 2002; Hayes, Walton, Szomor, & Murrell, 2001). However, no studies to date have specifically investigated the effect of passive and active end-range (active vs. passive) measures on humeral rotational pattern and scapular position in overhead throwing athletes.

The aim of this study was to quantify the effects of the active or passive end-range determination on the external and internal rotation ROM, as well as in the scapular position, in overhead throwing athletes assessed in a sitting position.

METHODS: A sample of 22 healthy subjects recruited from the community participated in this study and were divided in two groups: the athletes group (N= 11; age = 25.5 ± 5.9 years; height = 185.3 ± 7.9 cm; weight = 84.2 ± 9.3 kg) and the non-athletes group (N= 11; age = 27.4 ± 5.4 years; height = 172.7 ± 8.8 cm; weight = 73.3 ± 13.3 kg). Inclusion criteria for the athletes group was practicing overhead sports for at least 6 years. Non-athletes group included subjects that do not practice or have practiced overhead sports and do not have overhead professional activity. Subjects with a previous history of shoulder surgery or traumatic injury (e.g. dislocation, subluxation) were excluded from this study, as well as, participants with shoulder or elbow pain in the last 6 months. In a supine position with the dominant arm abducted at 90º, subjects were instructed to perform both active and passive shoulder rotation to establish the maximum range of humeral axial rotation. No allowance for scapular protraction or elevation was permitted. The scapulothoracic joint was stabilized via a posterior directed constainment force exerced by the examiner hand on the coracoid process and the anterior aspect of the acromion. This procedure replicates the one used on standard goniometry for shoulder rotation. Humeral and scapular 3D kinematic were recorded by means of an electromagnetic tracking device (Flock-of-Birds, Ascension Technology, Burlington, VT) controlled by a specific software (The Motion Monitor software, Innovative Sports Training, Chicago, IL) with a four sensors setup: the thorax sensor, firmly attached to skin over the first thoracic vertebrae (T1); the arm sensor attached by mean of a cuff just below the deltoid attachment; and the scapular sensor placed on the superior flat surface of the acromion process. A fourth sensor mounted on a hand-held stylus (±6.5cm) was used on bony landmarks digitalization in order to link sensors to the local anatomical coordinate systems (LCS) and subsequently calculated segments and joint rotations by combining the LCSs with the sensor motions. Segments LCSs and joint rotations definition were made according to the shoulder International Society of Biomecahnics and the International Shoulder Group standardization protocol (Wu, et al., 2005). The digitalization protocol was performed with the subject in a seated position, arm elevated (±90º), elbow flexed (±90º) and forearm parallel to the floor. This position was used on the definition of the neutral arm rotation position and the zero point (0º). The amplitude of arm rotation (internal or external) corresponds to the absolute value of the difference between this position and the end-range arm rotation position. Dependent variables includes humeral positions with respect to thorax (thoracohumeral angles) and to scapula (glenohumeral angles) as well as the 3D scapular position (protraction, lateral rotation and spinal tilt), recorded at the end-range of arm internal and external amplitude. A two-way repeated-measures ANOVA was used to calculate the effects of the end-range determination (passive or active), and arm rotation (internal and external) across groups (athletes and non-athletes) on dependent variables. Significant results were considered for p values < 0.05.

RESULTS: With respect to the internal rotation (IR), no differences were found between groups for all humeral and scapular variables. The active internal thoracohumeral (TH) and glenohumeral (GH) angles were significantly (p=0.00) higher than internal passive TH and GH. Concerning external rotation (ER), no differences were found between groups for all humeral variables but for scapular variables athletes showed less scapular posterior tilt (p=0.00) and a scapula more in retraction (p=0.027). That means a scapular position with the inferior angle of the scapula fairway from the thorax cage and simultaneously with the glenoid more oriented with the frontal plane.
External passive TH (33.7º ± 3.9º) was significantly higher (p=0.034) than external active TH (31.3 ± 4.1). On the same way external passive GH (34.5º ± 3.7º) was significantly higher (p=0.00) than external active GH (28.2º ± 4.0º).

DISCUSSION: Our findings showed that shoulder internal active ROM has higher values than passive motion. These results emphasize the importance of the end-range determination in a clinical setting, particularly on functional assessment of the thrower´s shoulder. Reports are inconsistent with regard to how end-range is determined. Some use active positioning while others use passive positioning determining capsular end-feel (Awan, et al., 2002; Barlow, et al., 2002; Reagan, et al., 2002), by scapular liftoff (Warner, et al., 1990) or by pain (Andrews & Bohannon, 1989). This aspect is crucial to understand the results from other studies that showed higher values of ROM associated to passive condition of testing (Myers, et al., 2006; Osbahr, Cannon, & Speer, 2002). Most of the studies in the literature assessed shoulder rotational ROM in supine position, and the arm at 90º abduction, like we did. Athletes presented higher internal rotation values within active motion. Concerning external rotation, passive motion showed highest values especially among non-athletes. We also found that athletes have more internal rotation than non-athletes passively or actively which is different from what we have found in literature (Dwelly, Tripp, Tripp, Eberman, & Gorin, 2009; Torres & Gomes, 2009). Considering GH, we found more active internal rotation, highest values among the athletes, but concerning external rotation passive motion is higher, and athletes are the ones that show the highest values. This could be due to shoulder osseous or soft-tissue adaptations that can result from repetitive shoulder motions (Huffman, et al., 2006; McCully, Kumar, Lazarus, & Karduna, 2005), which are common among throwing athletes. Stretching of the anterior glenohumeral capsule leads to increased external rotation at the point of late cocking and early acceleration and aids in the achievement of higher throwing velocities. Although literature (McCully, et al., 2005) refers that many throwers develop a posterior capsular contracture that limits internal rotation. We found in the athletes group more active internal rotation than on non-athletes group. Besides this no differences were found in the total arch of arm movement, even on active or passive one. Concerning scapular position at the end-range of active arm rotation, significant differences were found on scapular tilt between groups, in such a way that non-athletes showed a scapula more in a posterior tilt position. This can be due to the fact that athletes use their scapula along the throwing motion, and not only the at glenohumeral joint as the experimental setup imposed. Borich et al. (2006) found in athletes with impingement and IR deficit a greater scapular anterior tilt. In contrast we found less IR among athletes but scapular posterior tilt, although non-athletes show more shoulder internal rotation, and greater scapular posterior tilt. Besides differences found, athletes seem to show a similar behaviour in both studies.

CONCLUSION: Our findings emphasize the importance of the end-range determination in a clinical setting particularly on functional assessment of the throwers shoulder. On shoulder internal rotation the highest values of TH and GH internal angles were found when the end-range was actively determined. In contrast, the active end-range determination was associated with the lowest values of TH and GH on shoulder external rotation. No differences were found between athletes and non-athletes for all variables at internal rotation, but at external rotation athletes showed a scapula less in protraction and less in scapular posterior tilt.

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EFFECT OF PERFORMANCE FEEDBACK DURING 6 WEEKS OF VELOCITY BASED SQUAT JUMP TRAINING

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This study investigated the effect of instantaneous performance feedback (peak velocity) provided after each repetition of squat jump exercises in 13 professional rugby players. Players were randomly assigned to a feedback or non feedback group and completed three training sessions per week for six weeks. The relative magnitude (effect size) of the training effects for all performance tests were found to be small, except for 30m sprint which was moderate. The use of feedback was found to be possibly beneficial to increasing vertical jump, 10m and 20m sprint, likely to be beneficial to increasing horizontal jump and almost certainly beneficial to increasing 30m sprint. It is suggested that the provision of instantaneous feedback on movement velocity during resistance training sessions provides a greater potential for adaptation and larger training effects.

KEYWORDS: Monitor, resistance training, squat jumps, sprint.

INTRODUCTION: Although the monitoring of training load and or training intensity may provide useful information as to what has been completed in training, its value in affecting positive changes within a session or to quantify and evaluate each session is limited. It is fairly conclusive from motor learning theory that instantaneous feedback in terms of knowledge of performance and knowledge of results can have a substantial effect on athletic performance and the acquisition of motor skills (Bilodeau, 1966; Kilduski and Rice, 2003). Of particular interest is the literature citing improvements in strength and the acute production of force and power (6-12 %) when the subjects were given visual feedback (Figoni and Morris, 1984; Graves and James, 1990; Kellis and Baltzopoulos, 1996). However, the effects of this type of feedback over an entire resistance training cycle are unexplored and provide exciting possibilities for improved athletic performance.

One area in which this feedback might be most useful is in how the load is actually moved. Maximum power output is the product of optimum force and shortening velocity (Fleck and Kraemer, 2004; Zink *et al.*, 2006), therefore when training for power development it would seem intuitive to ensure movement velocity, force and/or power output is maximized for each repetition of an exercise session. Consequently, it would seem logical to monitor and provide feedback for these variables. The purpose of the present study was to investigate the effect of instantaneous performance feedback (peak velocity) provided after each repetition of squat jump exercises over a six week training block on sport specific jumping and sprinting performance tests.

METHOD: Thirteen professional rugby players were randomly assigned to one of two groups, feedback (n = 7, age = 25.7 ± 3.6 years, height = 188.5 ± 8.2 cm, mass = 104.3 ± 10.0 kg, training age = 3.7 ± 1.0 years, 1RM squat = 176.0 ± 35.6 kg) and non feedback (n = 6, age = 24.2 ± 2.5 years, height = 184.7 ± 7.2 cm, mass = 102.9 ± 14.3 kg, training age = 3.2 ± 1.2 years, 1RM squat = 185.4 ± 28.8 kg). All subjects had a minimum of two years resistance training experience and were currently in the pre-season phase of their training.

Each group completed a testing sessions at least 48 hours prior to the commencement of the training study and 48 hours after the completion of training. The testing sessions consisted of bilateral countermovement vertical and horizontal jumps, and 30m timed sprints with split times also taken at 10m and 20m. Three resistances sessions per week were prescribed and all participants completed the same exercises and number of repetitions and sets. All other conditioning sessions (energetic and skills focus) were similar for both groups of players.

Three sets of three concentric squat jumps were performed in two of the three sessions each week with a 40kg barbell. The depth of the squat was set at a knee angle of 90^{o}, controlled via an adjustable rack that the barbell rested upon prior to each repetition. Participants were

instructed to perform the movement as fast / explosively as possible with a pause between repetitions to distinguish each movement. The subjects in group one (feedback) were given real-time feedback (visual onto a screen) on peak velocity at the completion of each repetition, whilst those in group two (non feedback) did not receive any feedback. Peak velocity during the concentric phase for each repetition was recorded using a position transducer (Celesco PT5A-150; Chatsworth, CA) with a velocity repeatability of better than ± 0.10% of output, and customized data acquisition and analysis software (Labview, National Instruments, Austin TX). Velocity was calculated by differentiating the displacement time data which was sampled at 500 Hz and low-pass filtered at 10 Hz.
Intraclass correlation coefficients (ICC) were used to determine the consistency of effort (i.e. consistency of session average peak velocity) for both groups over the entire training study. A spreadsheet for analysis of a straight forward controlled trial (Hopkins, 2003) was used to determine the percent change between pre and post training study for each of the performance tests. Cohen effect sizes (ES) were used to determine the relative magnitude of the training effects. (ES < 0.41 represented a small ES, 0.41 to 0.70 a moderate ES, and > 0.70 a large ES (Cohen, 1988). The chances (% and qualitative) that the true value of the statistic (percent change in variable of interest) was practically or mechanistically positive, trivial, or negative was also calculated using the spreadsheet. An alpha level of 0.05 was also used for statistical significance.

RESULTS: The change in 30m sprint time was the only statistically significant difference between training groups (p = 0.0008). The mean (± SD) results and percent change of the performance test for the feedback and non feedback conditions can be observed in Table 1. These show that for all tests the feedback condition produced larger percent changes in means (0.9 to 4.6% vs. -0.3 to 2.8%).

Table 1 Mean, standard deviation (SD), and percent change in mean of vertical jump (m), horizontal jump (m), and 10/20/30 m sprints (s) pre and post six week squat jump training.

		Feedback	Non-Feedback
Vertical Jump	Pre	0.61 (0.06)	0.66 (0.06)
	Post	0.64 (0.07)	0.67 (0.01)
	Percent Change	4.6	2.8
Horizontal Jump	Pre	2.50 (0.16	2.58 (0.20)
	Post	2.56 (0.15)	2.59 (0.20)
	Percent Change	2.6	0.5
10 m Sprint	Pre	1.74 (0.04)	1.79 (0.10)
	Post	1.73 (0.05)	1.79 (0.09)
	Percent Change	-1.3	-0.1
20 m Sprint	Pre	3.03 (0.06)	3.06 (0.16)
	Post	3.00 (0.06)	3.06 (0.15)
	Percent Change	-0.9	-0.1
30 m Sprint	Pre	4.20 (0.11)	4.25 (0.21)
	Post	4.14 (0.11)	4.26 (0.19)
	Percent Change	-1.4	+0.3

With regards to practical significance, the chance that these changes were practically beneficial or trivial as well as the effect sizes are reported in Table 2. The use of feedback during squat jump training was reported to be possibly (45-65%) beneficial to increasing vertical jump, 10 m and 20 m sprint performance, likely (83%) to be beneficial to increasing horizontal jump performance and almost certainly (99%) beneficial to increasing 30 m performance. The relative magnitude (ES) of the training effects for all performance tests was small (0.18 to 0.28), except for the 30 m sprint performance which was moderate (0.46).
The ICC was used as a measure of consistency of effort between days. The ICCs for the feedback condition (0.81 to 0.95) were superior to the non-feedback condition (-0.52 to 0.14)

suggesting that those in the feedback group maintained effort/rank (i.e. average system velocity) to better effect than the non-feedback group.

Table 2 Effect sizes and chances (% and qualitative) that the benefit of feedback during jumps squats is practically positive or trivial for vertical jump, horizontal jump, and 10/20/30 m sprints following six weeks of training.

	Vertical Jump	Horizontal Jump	10 m Sprint	20 m Sprint	30 m Sprint
Effect Size	0.18 Small	0.28 Small	-0.28 Small	-0.20 Small	-0.46 Moderate
Positive	45 Possibly	83 Likely	65 Possibly	49 Possibly	99 Almost Certainly
Trivial	51	17	33	49	1

DISCUSSION: Results indicated an increase in vertical jump over the 6 weeks for both the feedback (4.6%) and non feedback (2.8%) groups. Although a greater improvement was seen with feedback there was a 51% chance this was trivial and 45% chance of being positive. This suggests there is some evidence for the use of feedback during training to enhance vertical jump performance. Given this performance test was very similar to the movement used in training (squat jump) it suggests that improvements were seen as a result of repetition of the movement regardless of the feedback conditions.

A larger increase in performance with the use of feedback was also observed in the horizontal jump (2.6% vs. 0.5%). As suggested previously it is thought that movements requiring a powerful thrust from hips and thighs can be improved through the prescription of a biomechanically similar movement during training (Adams et al., 1992). It would seem that this has occurred here where the use of squat jumps during training resulted in improvements in horizontal jump performance. Again there appears justification for the use of feedback within training to optimise performance improvements, as the use of feedback was reported as being likely to be beneficial to increasing horizontal jump performance (83% chance of a positive effect) and a small training effect noted (ES = 0.28).

Improvements in sprinting speed for the feedback group were observed over 10 m (1.3%), 20 m (0.9%) and 30 m (1.4%) distances. Again these were larger than those observed from the non-feedback group (0.1%, 0.1% and -0.3% respectively). This meant that feedback was possibly beneficial to increasing 10 m and 20 m sprint performance, with small training effects (ES = -0.28 and -0.20 respectively) and almost certainly beneficial to increasing 30 m performance, with a moderate training effect (ES =0.46). The results from the non-feedback group are in agreement with previous research using jumps without feedback, whereby loads of 70%1RM (Hoffman et al., 2005) and 30% (Wilson *et al.*, 1993) did not produce any significant increases in sprinting speed.

With regards to the motivational aspects of feedback it seems that the feedback resulted in a greater consistency of effort/performance throughout the programme as highlighted by the reported ICC values. The feedback group's ICCs ranged from 0.81 to 0.95 whereas the non-feedback condition ICs ranged from -0.52 to 0.14. Given the ICCs relate to the reproducibility of the rank order of subjects on a subsequent training session, it appears that the use of feedback during training enabled a greater consistency in the peak velocity achieved during the squat jumps. As it has been suggested that the actual velocity of training is a vital component of producing high velocities in other sporting movements (McBride et al., 2002), such a result appears of considerable importance. In addition peak velocity during traditional squats has been shown to be significantly correlated to sprint time ($r = 0.40$, $P = 0.029$) (Sleivert and Taingahue, 2004). Similarly it has also been suggested that exercises with greater rate of force development (RFD) lead to greater improvements in sprinting (Tricoli *et al.*, 2005), and whilst RFD was not measured in the present study consistently higher peak bar velocities were seen with feedback. Therefore it would appear that optimising the training session through the use of feedback leads to increases in sprint performance that may not have been realised using traditional training strategies.

CONCLUSION: Results of this study indicated that the provision of feedback on a single exercise (squat jump) during a resistance strength training programme resulted in an improvement in the performance of movement and sport specific tests. Given athletes were also able to produce more consistent training performances through the entire six week training programme, it would seem intuitive to constantly monitor multiple exercises of each training session and provide feedback, which should provide greater potential for adaptation and larger training effects. The use of such monitoring and feedback technologies may be further utilised through the ability to set training performance targets, such as maximum velocity and number of repetitions and/or sets completed above a pre determined performance threshold. This may prove to be very motivational when fatigue sets in, as well as creating competition between athletes in the training environment.

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Poster Session 3

THE TIME COURSE OF RECOVERY FROM A MESOCYCLE OF PERIODIZED PLYOMETRIC TRAINING

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This study evaluated the effectiveness of a mesocycle of periodized plyometric training and the influence of the duration of the post training recovery period. All subjects' countermovement jump height, peak power, and body mass were assessed with a force platform prior to and 2, 4, 6, 8, and 10 days after training. Jump height was 25.0% greater ($p \le 0.05$) after training with no difference ($p > 0.05$) between the recovery periods of 2, 4, 6, 8, or 10 days. Peak power was 11.6 to 14.3 % greater ($p \le 0.001$) after training for the training group with no difference ($p > 0.05$) between recovery periods of 2, 4, 6, 8, or 10 days. Periodized plyometric programs with decreasing volume and increasing intensity improve jump performance without a need for a post training recovery period.

KEYWORDS: jump training, stretch shortening cycle, taper, rest, fatigue

INTRODUCTION: The positive effect of plyometric training on jumping performance has been well established in the literature (Markovic, 2007). However, the specifics of plyometric program design remain unclear. The systematic use of periodization, as well as the related concepts of the training taper (Bosquet et al., 2007) and post training recovery period (Weis et al., 2003), is well established for some training modalities such as strength training and may also be applied to plyometric program design.

Key features of periodized programs such as a systematic decrease in volume or increase in exercise intensity are not used in many plyometric training studies (Markovic, 2007). Nonetheless, some studies demonstrate small to moderate improvement in countermovement jump height (Chimera, et al., 2004; Fatouros et al., 2000; Gehri, et al., 1998) and power (Fatouros et al., 2000). In some cases, countermovement jump height and power did not improve, or even decreased, when testing was performed immediately after training, and only improved after a period of recovery (Luebbers et al., 2003). Thus, recovery from the plyometric training stimuli seems important.

Popular literature includes recommendations for the increase in plyometric intensity and decrease volume (Potach and Chu, 2008), though the specifics for doing so remain unclear. Previous plyometric research has begun to quantify the intensity of plyometric exercises (Ebben et al., 2008; Jensen and Ebben, 2007) and recommendations have been made for the development of periodized plyometric programs (Jensen and Ebben, 2007).

The training taper prior to competition is related to periodization in that each share the goal of reducing training volume in order to maximize performance. Performance of a variety of exercise modes may be optimized with a 41-60 percent reduction in training volume (Bosquet et al., 2007). Two studies specifically compared a no training recovery period to a period of reduced volume taper demonstrating superior performance in torque, strength, and power with tapering than with a non training recovery period of 10 days (Gibala, et al., 1994) or 4 weeks (Izquierdo et al., 2007).

To date, the application of periodization to plyometric training programs and the value of periodization and/or post training recovery has not been investigated. The purpose of this study is to evaluate the effect of a mesocycle of periodized plyometric program and the duration of the post training recovery period that optimizes jump height and peak power during the countermovement jump.

METHODS: Fourteen women served as training subjects (mean ± SD, age 19.29 ± 0.91 yr; body mass 62.56 ± 7.24 kg; height 167.19 ± 6.51 cm). Controls included 10 women (mean ± SD, age 19.5 ± 1.18 yr; body mass 60.41 ± 7.93 kg; height 163.45 ± 6.50 cm). Body mass was assessed for all test sessions and a repeated measures ANOVA showed no change for the training or control groups across any of the test sessions as described in Table 1. The subjects were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board.

Prior to all testing and training sessions, subjects warmed up and performed dynamic stretching exercises and 5 countermovement jumps of increasing intensity. All training and control group subjects were instructed to refrain from physical activity during the 6 week training period which was confirmed via analysis of subject activity logs.

Subjects participated in a pre-training testing session and five post training testing sessions. The post training testing sessions were performed 2, 4, 6, 8, and 10 days after the 6 week training program for training subjects, and 6 weeks after the pre test for the control subjects. The pre training and post training testing sessions consisted of 3 repetitions of the countermovement jump.

Subjects were randomly assigned to either a non-training control or plyometric training group. The plyometric group trained twice per week with 48 to 96 hours recovery between training sessions. The program was periodized consistent with previous recommendation for decreasing volume and increasing plyometric intensity (Potach and Chu, 2008). The volume was reduced by 40 percent from a high of 100 foot contacts early in the program to 60 foot contacts near the end of the program. This degree of volume reduction is consistent with the results of a meta-analysis showing performance is optimized with this degree of training volume reduction (Bosquet et al., 2007). The total volume of the plyometric program was 475 foot contacts. The intensity of the plyometric exercises was determined based on previous research examining ground reaction forces, knee joint reaction forces, and muscle activation (Ebben et al., 2008; Jensen and Ebben, 2007). Subjects rested approximately 30 seconds between sets and 15 seconds between single jumps. The recovery duration between reps and sets was chosen based on previously recommended work to rest ratios of at least 1:5 (Potach and Chu, 2008), research showing that there is no advantage in jump performance with more than 15 seconds rest between repetitions (Read and Cisar, 2001).

The countermovement jump tests were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies Inc., Watertown, MA). The force platform was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed, and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA) for later analysis. Jump height and peak power were analyzed since these variables are frequently used to assess countermovement jump performance (Canavan and Vescovi, 2004, Moir, 2008). Jump height was calculated from the force-time records consistent with methods previously used (Moir, 2008). Peak power was calculated using the equation proffered by Canavan and Vescovi (2004).

Data were analyzed with SPSS 17.0 using a repeated measures ANOVA with Bonferroni adjusted pairwise comparison in order to identify the specific differences in jump height, peak power, and body mass between the pre-training baseline testing and testing sessions performed at 2, 4, 6, 8, and 10 days after training. The reliability of the trials was assessed using intraclass correlation coefficient (ICC), for each of the dependent variables for the pre-training and last post-training testing session. Assumptions for linearity of statistics were tested and met. Statistical power (d) and effect size (η_p^2) are reported and all data are expressed as means ± SD. The *a priori* alpha level was set at $p \leq 0.05$.

RESULTS: Results revealed significant main effects for countermovement jump height ($p \leq 0.001$, $d = 0.98$, $\eta_p^2 = 0.41$) and peak power ($p \leq 0.001$, $d = 1.00$, $\eta_p^2 = 0.56$), but not for body mass ($p > 0.05$), between test sessions, for the subjects in the plyometric training group. Post hoc analysis demonstrated that jump height and peak power were different between the pre-training testing session and all post training testing sessions, with no difference between any of the post training testing sessions. Results of post hoc analysis are shown in Table 1. No significant main effects were found, demonstrating no differences in countermovement jump height ($p > 0.05$), peak power ($p > 0.05$), or body mass ($p > 0.05$) between testing sessions, for subjects in the control group. Interclass correlation coefficients were calculated for the training subjects and controls for the countermovement jump height and power with all values ranging between 0.84 and 0.99.

Table 1. Training (N=14) and control (N=10) group jump height (cm) and power (W), each expressed as mean ± SD prior to training and 2, 4, 6, 8, 10 days post training

	Training Group		Control Group	
	Jump height	Power**	Jump Height	Power
Pre training	0.21 ± 0.08	1810.98 ± 323.76	0.25 ± 0.06	1861.23 ± 428.68
2 days post training	0.28 ± 0.03*	2053.13 ± 305.18**	0.25 ± 0.05	1874.41 ± 452.45
4 days post training	0.28 ± 0.03*	2048.80 ± 331.21**	0.25 ± 0.05	1929.07 ± 455.42
6 days post training	0.28 ± 0.03*	2073.83 ± 283.51**	0.25 ± 0.05	1910.06 ± 470.26
8 days post training	0.28 ± 0.04*	2088.16 ± 314.65**	0.25 ± 0.05	1938.12 ± 428.33
10 days post training	0.28 ± 0.04*	2076.36 ± 320.60**	0.25 ± 0.05	1931.24 ± 410.17

*Significantly different from the pre training value ($p \leq 0.05$)

**Significantly different from the pre training value ($p \leq 0.01$)

DISCUSSION: This study demonstrates that a mesocycle of periodized plyometric training produces substantial improvement in vertical jump height and peak power. The length of the post training recovery period does not influence jump performance, presumably due to the tapering inherent in periodized plyometric training. Thus, the performance of the subjects was optimal within 2 days of training and performance adaptations were sustained for at least 10 days after training.

This performance increases in this study were greater than those that demonstrated either no increase in countermovement jump height (Vescovi, et al., 2008) or increases that ranged from 2.8 to 10.2 % (Chimera et al., 2004, Fatouros et al., 2000, Gehri et al., 1998, Markovic et al., 2007). In the present study, the periodized program design including exercises of known increasing intensity (Ebben et al., 2008; Jensen and Ebben, 2007) and decreasing training volumes in the recommended range (Potach and Chu, 2008) were more optimal compared to other studies. Most other plyometric programs used no systematic increase in exercise intensity or decrease in volume. In fact, some studies included plyometric volumes that increased up to 480 foot contacts per session over the course of the training program (Chimera et al., 2004).

Results of this study confirm that tapered programs with a 41-60 % decline in volume, enhances performance (Bosquet et al., 2007). The present study showed that performance improved with no difference between recovery periods of 2, 4, 6, 8, or 10 days, indicating that periodized programs may peak athletes after training and prior to competition without the need for a post training recovery phase.

Previous research comparing a non-training recovery period to a tapering period of reduced volume demonstrated superior performance in strength and power measures after training with reduced volume tapering than with a non-training recovery period of 10 days (Gibala, et al., 1994) or 4 weeks (Izquierdo et al., 2007). Thus, results of the present study add to the body of literature indicating that systematic volume reduction, and not a non-exercising recovery period, may be more ideal for performance enhancement.

Results of the present study call into question the previously held belief that training programs should be longer than 10 weeks to be highly effective (De Villarreal et al., 2009).

CONCLUSION: The present study demonstrates that a brief, moderate volume periodized mesocycle of plyometric training produces large improvements in countermovement jump compared to the pretest performance, without the need for and regardless of the length of the recovery period at the end of the training cycle.

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EVALUATING PLYOMETRIC EXERCISE USING REACTIVE STRENGTH INDEX-MODIFIED

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The reactive strength index (RSI) measures explosive power during depth jumps. The purpose of this study was to introduce a modification of the RSI (RSI_{mod}) that can be used to evaluate the explosive power of any vertical plyometric exercise, to assess its reliability, and evaluate the RSI_{mod} of a variety of plyometric exercises. Forty nine subjects performed 3 repetitions of 5 plyometric exercises including the countermovement jump, tuck jump, single leg jump, squat jump, and dumbbell countermovement jump. Data were analyzed using a two way ANOVA. Results reveal significant differences in RSI_{mod} between all plyometric exercises ($P \leq 0.001$). The RSI_{mod} was highly reliable for all of the plyometric exercises studied. The RSI_{mod} offers a reliable method of assessing the explosiveness developed during a variety of plyometric exercises.

KEYWORDS: countermovement jump, instrumentation, athlete testing, power, reliability

INTRODUCTION: This ability to develop maximal force in a minimal amount of time is a requisite ability in most sports (Zatsiorsky and Kraemer, 2007). The reactive strength index (RSI) is a measure of force and the time it takes to develop that force, by calculating the jump height divided by ground contact time during the depth jump (Flanagan et al., 2008).
The RSI has been found to be a reliable scientific measure (Flanagan et al., 2008), a practical way to evaluate the quality of training of sports teams (McClymont, 2003), and a diagnostic test of functional ability for those with ACL reconstructed legs (Flanagan & Harrison, 2006). At present, the RSI is typically used to evaluate the performance of the depth jump since it is the only plyometric exercise with an identifiable ground contact time. However, most other plyometric exercises are initiated with a countermovement.
Recent research has assessed a variety of characteristics of plyometric exercises using electromyography (Ebben et al., 2008), and kinetic data such as ground and knee joint reaction forces (Jensen & Ebben, 2007). Previous research has examined the RSI based on the initiation of the countermovement and the duration of the eccentric and concentric phases of the stretch shortening cycle, rather than contact time, for bilateral and single leg countermovement jumps (Ebben et al., 2009). This concept has been also referred to as a ratio of the flight time during a jump to the contraction time (Cormack et al., 2008).
The evaluation of other plyometric variations using a measure similar to the RSI would be useful to assess these exercises characteristics. Therefore, the purpose was to assess a modification of the reactive strength index (RSI_{mod}) that can be applied to all vertical plyometrics using the time to takeoff, rather than ground contact time, divided by the jump height. This study also assessed the reliability of the RSI_{mod} for a variety of plyometric exercises and evaluated the intensity of these exercises based on the RSI_{mod}.

METHODS: Twenty-six men (20.23 ± 1.63 years) and 23 women (20.39 ± 1.50 years) served as subjects. All subjects were recreationally fit and participated in weekly resistance and plyometric training. The subjects were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board.
All subjects performed a habituation and testing session. Prior to each session, the subject warmed-up with 3 minutes of low intensity work on a cycle ergometer, performed dynamic stretching and 5 countermovement jumps of increasing intensity.

During the habituation session, the subjects' countermovement jump height and 5 repetition maximum (RM) back squat were assessed. Subjects were then given instruction and demonstration of the correct technique for the plyometric exercises to be assessed during the test session. Subjects performed each of these exercises until they mastered the technique. The plyometric exercises included the squat jump (SJ), tuck jump (TJ), countermovement jumps (CMJ), loaded countermovement jump with handheld dumbbells equal to 30% of the subjects previously assessed estimated 1 RM squat (DBJ), and right leg single leg jump (SLJ). The subject also performed a depth jump (DJ) with a box height normalized to their vertical jumping ability, consistent with methods previously used (Ebben et al, 2010). These exercises were performed based on previously described methods (Potach & Chu, 2008). These plyometric exercises were tested since they represent a variety of estimated and researched plyometric exercise intensities (Ebben et al., 2008; Ebben et al., 2009; Jensen & Ebben, 2007; Potach & Chu, 2008), and to allow the assessment of unilateral and bilateral and loaded and unloaded plyometric exercise conditions, and to assess those that use and do not use the stretch shortening cycle. The DJ was also performed in order to allow descriptive comparison of the conventional RSI value to the RSI_{mod} values obtained for the test plyometric exercises.

During the testing session subjects performed 3 repetitions of each of the test plyometric exercises in a randomized order with 1 minute of rest between each exercise. Randomization and adequate rest between sets based on previous recommendations (Potach & Chu, 2008, Read & Cisar, 2001) was used to control order and fatigue effects.

The test exercises were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies Inc., Watertown, MA, USA). The force platform was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Inc., Watertown, MA USA) for later analysis.

The RSI_{mod} was created based on the RSI. The RSI_{mod} replaces ground contact time with time to takeoff in the equation. Thus, jump height is divided by time to takeoff. Time to takeoff includes the eccentric and concentric phases of the stretch shortening cycle and can be calculated for all vertical plyometric exercises. In this study, time to takeoff was calculated from the force time record as the time of onset of the flight phase minus the time of onset of the eccentric phase or countermovement. The time of onset of the eccentric phase was identified consistent with methods previously described (Jensen et al., 2009). Jump height was calculated using previously published equations (Jensen et al., 2009; Moir, 2008). The DJ was also analyzed using conventional RSI calculations (Flanagan et al., 2008) to allow for descriptive comparison with the plyometric exercises evaluated using RSI_{mod}. All values were determined as the average of 3 trials of each plyometric exercise. Figure 1 shows a force-time record from a countermovement jump and provides an example of the time to takeoff and flight time, which were used to calculate jump height.

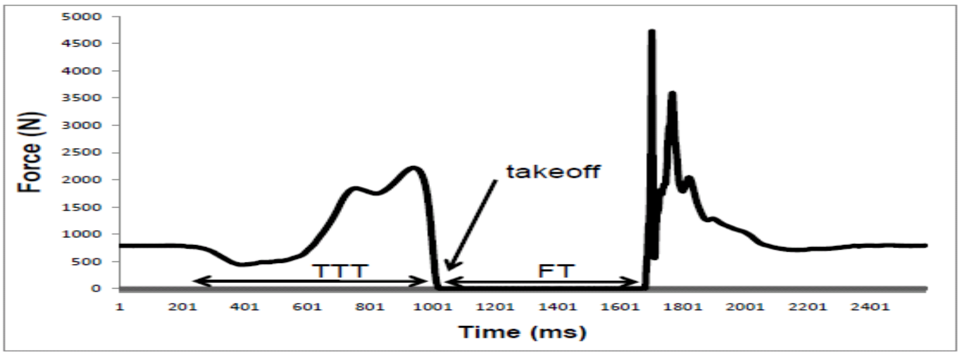

Figure 1. Force time record with dependent variables. takeoff, TTT = time to takeoff, FT = flight time

A repeated measures ANOVA was used to evaluate the main effects for $RSI_{mod.}$ Significant main effects were further analyzed with Bonferroni adjusted pairwise comparison in order to identify the specific differences in RSI_{mod} for the plyometric exercises assessed. The trial to trial reliability was assessed for the RSI_{mod} for each plyometric exercise using both single (ICCsingle) and average (ICCave) measures intra-class correlations. The ICC classifications of Fleiss (1986) (less than 0.4 was poor, between 0.4 and 0.75 was fair to good, and greater than 0.75 was excellent) were used to describe the range of ICC values. In addition, a repeated measures ANOVA was used to confirm that there was no significant difference in RSI_{mod} between three trials of each plyometric exercise. An *a priori* alpha level of $P \leq 0.05$ was used with post hoc power and effect size represented by d and η_p^2, respectively.

RESULTS: The analysis of RSI_{mod} revealed significant main effects for plyometric exercise type ($P \leq 0.001$, $\eta_p^2 = 0.79$, $d = 1.00$) but not for the interaction between plyometric exercise type and gender ($P > 0.05$). Results of Bonferroni adjusted pairwise comparisons are presented in Table 1 with mean data included for both men and women to provide normative RSI_{mod} data for each gender for the plyomeric exercises assessed. Intraclass correlation coefficients assessed for all plyometric exercises RSI_{mod} assessing the trial to trial reliability ranged from 0.85 to 0.90 for single measures and 0.95 to 0.96 for average measures, for all of the plyometric exercises assessed, with no significant differences found between three trials of each plyometric exercise ($P > 0.05$). The average RSI for the DJ for all subjects was 1.99 ± 0.28. This variable was not compared statistically to the RSI_{mod} since each was calculated with different equations.

Table 1. Reactive strength index modified expressed as a ratio of the jump height divided by the time to takeoff for the following plyometric exercises for men (N=26) and women (N=23). Data are presented as mean ± SD.

	Men	Women
Tuck Jump	0.81 ± 0.21	0.57 ± 0.14
Countermovement Jump	0.74 ± 0.19	0.52 ± 0.17
Squat Jump	0.71 ± 0.14	0.48 ± 0.10
Single Leg Jump	0.36 ± 0.12	0.24 ± 0.09
Dumbbell Countermovement Jump	0.31 ± 0.08	0.22 ± 0.04

DISCUSSION: This study introduces the concept of the RSI_{mod} to the literature, demonstrates its high level of reliability for a variety of plyometric exercises (Fleiss, 1986), and shows that there are differences in RSI_{mod} for all of the plyometric exercises assessed. The reliability of the RSI_{mod} is similar to the RSI based on a comparison of current results and previously reported ICC values (Flanagan et al., 2008). The RSI is typically calculated for the DJ and the value obtained may depend on the depth jump box height (McClymont, 2003). The RSI_{mod} would not be confounded by box height choice as the DJ is, since an athlete's performance of the eccentric and concentric phases of the stretch shortening cycle are likely to be fairly uniform for most plyometric exercises.

The DJ produced mean RSI values that were higher than the RSI_{mod} values of the plyometric exercises in the present study, most likely due to the limited contact time associated with the DJ compared to the higher time to takeoff values of the other plyometric exercises assessed. Subjects in the current study were specifically instructed to land from the DJ and quickly transition to the response jump, consistent with previous procedures (Flanagan et al., 2008). Previous reports demonstrate RSI values in a range of 1.29 to 1.70 with mean DJ contact times ranging from 225 to 274 ms, depending on DJ box height (McClymont, 2003). Thus, during the DJ, contact time representing the duration of the eccentric and concentric phase of the stretch shortening cycle was relatively brief (McClymont, 2003). In the present study, the DJ contact times used to calculate the RSI averaged approximately 200 ms. On the other hand, the time to takeoff values used to calculate the RSI_{mod} ranged from 558 ms for the CMJ to 781 ms for the SLJ, producing comparatively low RSI_{mod} values. This observation suggests the relative value of the DJ as a training strategy, though previous research

indicated its characteristics are dependent on box height (Ebben et al., 2008; Jensen & Ebben, 2007). Furthermore, since subjects are instructed to land and jump as fast and as high as possible for the DJ (Flanagan et al., 2008; McClymont, 2003), they should be taught a similar execution for all other plyometric exercises.
The RSI is a valuable measure since it measures reactive strength (Flanagan & Harrison, 2006; Flanagan et al., 2008; McClymont, 2003), assesses athlete performance in training (McClymont, 2003) and identifies injured athletes readiness for return to sport (Flanagan & Harrison, 2006). It is possible that the RSI_{mod} can serve many of the same functions while allowing the assessment of many more variations of plyometric exercises in the process.

CONCLUSION: Results of this study indicate that the $RSI_{mod,}$ determined by using flight time divided by time to takeoff, is a reliable method of assessing explosive strength for a variety of plyometric exercises. Men and women respond similarly to the RSI_{mod}. Plyometric exercises with the highest RSI_{mod}, such as the TJ are the best for developing explosive strength. Plyometric exercise intensity can be increased throughout a program based on progressively incorporating exercises with higher RSI_{mod} over time.

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THE EFFECT OF CONCURRENT ACTIVATION POTENTIATION ON THE KNEE EXTENSOR AND FLEXOR PERFORMANCE OF MEN AND WOMEN

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This study evaluated the effect of remote voluntary contractions (RVC) during isometric and isokinetic knee flexion and extension tests and evaluated gender differences therein. Subject peak torque, rate of torque development, and power were assessed with a dynamometer in RVC and no RVC's conditions. A two way mixed ANOVA with repeated measures for condition was used to evaluate the interaction between conditions and gender, and to assess the main effects. Main effects were evaluated with a paired samples *t*-test. Results revealed a significant interaction between all but one test condition and gender as well as significant main effects for all of the variables assessed ($P \leq 0.05$). Men attained 9.2% to 19.7% greater performances in the RVC condition for all variables whilst women demonstrated no significant differences between test conditions.
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KEYWORDS: remote voluntary contractions, ergogenic, strength, power

INTRODUCTION: Maximizing the magnitude and rate of muscular force production is the goal of many athletic training programs. As a result, ergogenic training strategies have been sought. The concept of concurrent activation potentiation (CAP) has been proposed to capitalize on the simultaneous contractions of muscles remote to the prime mover, potentially augmenting the performance of the prime mover via motor overflow (Ebben, 2006). Recent research has yielded some evidence describing potential advantages of CAP (Ebben et al., 2008a; Ebben et al., 2008b). Nonetheless, this phenomenon and potential gender differences in response to CAP have yet to be assessed.

Recent evidence supports the idea that RVC's influence motor performance. Research evaluating the effect of jaw clenching on rate of force development (RFD) demonstrated that clenching the jaw before, as well as before and during a test of grip strength, resulted in 8.5% and 15.8 % greater RFD, respectively, compared to when the jaw was not clenched (Hiroshi, 2003). Others have tested the effects of RVC's such as jaw clenching during an athletic task, demonstrating that athletes manifested 19.5% higher RFD during the counter movement jump with a clenched jaw, compared to without jaw clenching (Ebben et al., 2008a). Thus, jaw clenching increases RFD in both clinical and athletic performance tests. A variety of RVC's in addition to jaw clenching have been proposed to be potentially effective in stimulating CAP (Ebben, 2006) and an aggregate of jaw clenching, hand gripping, and the Valsalva maneuver was found to be more effective than RVC conditions such as jaw clenching or hand gripping alone (Ebben, et al., 2008b).

Research demonstrating an increased RFD during the countermovement jump when jaw clenching was used as a RVC, employed both men and women as subjects, but did not specifically assess gender differences. Other studies assessed the CAP effect using only men as subjects. At present, research examining the effect of CAP is limited to studies assessing RFD during gripping and jumping, and isometric tests of torque development during isometric knee extension with male subjects. Additional research examining the effectiveness of RVC's is warranted and potential gender differences in response to this phenomenon need to be investigated. Therefore, the purpose of this study was to investigate the effect of RVC's on peak torque, rate of torque development, power, and work, in both isometric and dynamic conditions and to assess gender differences therein.

METHODS: Subjects included 11 men (mean ± SD, age 21.63 ± 1.80 yr; body mass 83.68 ± 10.40 kg) and 10 women (mean ± SD, age 20.70 ± 1.34 yr; body mass 66.36 ± 5.11 kg) who participated in collegiate athletics, club sports, or intramural sports. Baseline isometric (ISOM) knee extension peak torque and isokinetic (ISOK) knee extension and flexion peak torque in the NO-RVC condition revealed that women in this study had approximately 71.5 to 73.2% of the ability of their male counterparts. The subjects were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board.
After general and dynamic warm up, subjects were positioned on and strapped in a dynamometer (System 4, Biodex Inc., Shirley, NY) according to manufacturer specifications. The knee was positioned goniometrically at 90° and calibrated with the system software. During the ISOM condition, the knee joint angle was adjusted until the software indicated the knee was at 60° of flexion. The ISOK condition was performed at $60° \cdot sec^{-1}$.
Subjects then performed test specific warm up sets of leg extension and flexion exercises in two test conditions. One condition included performing the exercises with an open mouth and pursed lips, thus limiting the likelihood of jaw clenching, and consistent cycling between inspiratory and expiratory flow in order to reduce the Valsalva effect (NO-RVC). In this condition, subjects also held hand dynamometers (Lafayette Hand Dynamometer, model 78010, Lafayette Industries, Lafayette, IN) which were used to confirm the absence of hand gripping. The RVC condition included maximal jaw clenching on a dental vinyl mouth guard, the performance of the Valsalva maneuver, and maximal bilateral hand gripping using hand dynamometers. Subjects performed test specific warm ups in both NO-RVC and RVC conditions at 75%, and at 100% of their self perceived maximum ability.
The test consisted of subjects performing 1 ISOM knee extension for 5 seconds in both the NO-RVC and RVC conditions at 100% of their maximum ability, as well as 1 set of 3 repetitions of ISOK knee extension and flexion in both the NO-RVC and RVC conditions, at 100% of their self perceived maximum ability. All test sets were counterbalanced and randomized with 4 minutes of recovery between tests to reduce order and fatigue effects.
Torque curves for each subject were analyzed using manufacturer's software. Data were sampled for seconds 2-4 of the 5 second ISOM test exercises. Peak torque and rate of torque development were calculated for each 3 second sample in the ISOM condition. Peak torque, rate of torque development, and power, were calculated for knee flexion and extension using the average values obtained from all 3 repetitions in the ISOK condition. Rates of torque development were calculated for the first 300 ms of each test exercise and normalized to a second, for both the ISOM and ISOK conditions.
All data were analyzed using SPSS 16.0. A two way mixed ANOVA with repeated measures for condition was used to evaluate the interaction between NO-RVC and RVC conditions and gender, and to assess the main effects. Significant main effects were further evaluated with a paired samples *t*-test. Additionally, gender differences in strength and ISOM and ISOK torque in the NO-RVC condition were assessed with an independent samples *t*-test. Assumptions for linearity of statistics were tested and met. Statistical power (d) and effect size (η^2) are reported and all data are expressed as means ± SD. The *a priori* alpha level was set at $P \leq 0.05$.

RESULTS: A significant interaction was found between test condition and gender for ISOM peak torque, rate of torque development, ISOK knee flexion peak torque, ISOK knee extension rate of torque development, ISOK knee flexion rate of torque development, ISOK knee extension power, and ISOK knee flexion power ($P \leq 0.05$ for all interactions). Thus, differences exist in the response of men and women to the RVC and NO-RVC test conditions.
Analysis revealed significant main effects for ISOM peak torque for condition ($P = 0.006$, $d = 0.83$, $\eta^2 = 0.33$) and gender ($P = 0.001$, $d = 1.00$, $\eta^2 = 0.95$), and ISOM rate of torque development for gender ($P = 0.009$, $d = 0.79$, $\eta^2 = 0.31$), but not condition ($P = 0.09$, $d = 0.40$, $\eta^2 = 0.14$). The specific responses of men and women in the RVC and NO-RVC

conditions were evaluated with tests of simple effects for all ISOM tests conditions and gender with results described in Table 1.

Table 1. Mean peak torque and rate of torque development (± SD) during ISOM knee extension.

	Men (N=11)			Women (N=10)		
	NO-RVC	RVC	%	NO-RVC	RVC	%
PT (N·m)	227.51 (61.88)	253.21 (66.41)	10.16*	164.65 (20.51)	166.09 (30.95)	0.87
RTD ($N·sec^{-1}$)	323.00 (95.01)	402.32 (128.57)	19.72†	251.92 (83.62)	239.89 (87.17)	-5.00

PT = peak torque; RTD = rate of torque development; % represents the percentage difference between conditions when the NO-RVC is divided by the RVC condition
* Significant difference between NO-RVC and RVC conditions ($P < 0.01$)
† Significant difference between NO-RVC and RVC conditions ($P < 0.05$)

Analysis yielded significant main effects for ISOK knee extension peak torque for condition ($P = 0.001$, $d = 0.96$, $\eta^2 = 0.44$) and gender ($P \leq 0.001$, $d = 1.00$, $\eta^2 = 0.96$), and ISOK knee extension for condition ($P = 0.004$, $d = 0.88$, $\eta^2 = 0.37$), and gender ($P \leq 0.001$, $d = 1.00$, $\eta^2 = 0.98$). Main effects were found for ISOK knee extension rate of torque development with significant main effects for condition ($P = 0.006$, $d = 0.83$, $\eta^2 = 0.34$) and gender ($P = 0.004$, $d = 1.00$, $\eta^2 = 0.93$) as well as for ISOK knee flexion rate of torque development for condition ($P = 0.001$, $d = 0.95$, $\eta^2 = 0.44$), and gender ($P \leq 0.001$, $d = 1.00$, $\eta^2 = 0.96$). The analysis of main effects for ISOK knee extension power, resulted in significant main effects for condition ($P \leq 0.001$, $d = 0.99$, $\eta^2 = 0.52$) and gender ($P \leq 0.001$, $d = 1.00$, $\eta^2 = 0.96$). Similarly, for ISOK knee flexion power, significant main effects were seen for condition ($P \leq 0.001$, $d = 0.98$, $\eta^2 = 0.50$), and gender ($P \leq 0.001$, $d = 1.00$, $\eta^2 = 0.97$). In order to evaluate the main effects and the specific responses of men and women in the NO-RVC and RVC conditions, simple effects for all ISOK test conditions and gender were evaluated with results depicted in Table 2.

Table 2. Three repetition average (± SD) peak torque, rate of torque development, power, and work during ISOK knee extension (KE) and knee flexion (KF).

	Men (N=11)			Women (N=10)		
	NO-RVC	RVC	%	NO-RVC	RVC	%
KE PT (N·m)	218.40 (44.39)	251.42 (60.27)	13.13*	159.91 (24.46)	173.80 (38.94)	8.00
KE RTD ($N·sec^{-1}$)	419.12 (109.40)	480.39 (143.6)	12.76**	298.53 (68.49)	298.58 (93.39)	0.00
KE Power (W)	158.92 (35.46)	180.84 (42.89)	12.13**	111.99 (16.45)	115.88 (21.48)	3.36
KF PT (N·m)	116.91 (15.14)	136.46 (27.52)	14.33*	83.55 (12.14)	85.82 (15.55)	2.66
KF RTD ($N·sec^{-1}$)	225.73 (41.54)	259.69 (35.93)	13.08*	155.79 (39.91)	158.05 (54.28)	1.53
KF Power (W)	90.74 (15.56)	103.11 (19.68)	12.00*	62.75 (8.95)	64.98 (13.29)	3.44

KE = knee extension; KF = knee flexion; PT = peak torque; RTD = rate of torque development; % represents the percentage difference between conditions when the NO-RVC is divided by the RVC condition
* Significant difference between NO-RVC and RVC conditions ($P < 0.01$)

DISCUSSION: This is the first study to investigate gender differences in CAP. Men demonstrated statistically higher performances of 9.2 to 19.7% in the RVC compared to the NO-RVC condition, for all outcome variables assessed. In contrast, women showed no significant differences between conditions for any of the outcome variables. The results of this study confirm the ergogenic effect of CAP for men, for the outcome variables assessed, consistent with previous recommendations (Ebben, 2006) and research examining isometric measures (Ebben et al., 2008b).

For the men in the present study, peak torque was 10.2% greater in the RVC compared to the NO-RVC condition which was similar to past research that revealed isometric mean and peak knee extensor torque was approximately 14.6 and 14.8% greater, respectively, in the RVC compared to NO-RVC condition (Ebben et al., 2008b). In the present study, an aggregate of jaw clenching, handgripping, and the Valsalva maneuver was used since it has been demonstrated to be the most effective RVC condition (Ebben et al., 2008b). However, unlike previous research (Ebben et al., 2008b) the current study examined and found gender differences.

Past work in this area revealed that compared to the non-jaw clenching condition, jaw clenching increased RFD during the countermovement jump by 19.5% (Ebben, et al., 2008a). This range of RFD augmentation compares similarly to the increased rate of torque development demonstrated by the men in the RVC condition in the present study which was 19.7% during isometric testing, and 12.7 and 13.1% during isokinetic knee flexion and extension testing, respectively.

The women subjects in this study demonstrated no statistically greater performance in the RVC compared to the NO-RVC condition for any of the outcome variables. This finding is without precedent in the limited literature examining CAP. However, in a study designed to assess the effect of using dynamometer handgrips on subject stability during isokinetic testing, gripping the dynamometers handles resulted in a statistically significant difference of 8.4% in knee extensor torque, compared to the condition where subjects crossed their arms over their upper torso (Stumbo et al., 2001). It is interesting to note that in this study, only the performance of men was greater in the handgripping condition (Stumbo et al. 2001). While increased stability may have augmented the performance of the male subjects only, it is also possible that this study demonstrates an unintended CAP effect as a result of motor overflow that is specific to men only (Stumbo et al., 2001), consistent with the results of the present study.

Baseline tests of NO-RVC ISOM and ISOK extension and flexion torques revealed values for women that were 71.5 to 73.2% of their male counterparts. Thus, the women in this study appear to be well trained with respect to the men and larger than typical gender differences in strength are not an explanation for the gender differences found in this study.

A review of other potentiation phenomenon, such as post-activation potentiation demonstrates no significant effect in many studies and performance enhancement ranging from 2.0 to 9.5% for the studies that demonstrated an effect. (Hodgson et al., 2005). In contrast, the men in the present study performed in a range that was 9.2 to 19.7% higher in the RVC compared to the NO-RVC condition.

CONCLUSION: Results from the present study expand the findings of an ergogenic effect of CAP in men for every variable assessed. Remote voluntary contracts have potential to augment the performance of strength training exercises.

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THE EFFECT OF REMOTE VOLUNTARY CONTRACTIONS DURING FAST STRETCH SHORTENING CYCLE ACTIVITY

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This study evaluated the effect of remote voluntary contractions (RVC) on depth jump performance. Subjects performed the depth jump in a RVC condition and a condition without RVC (NO-RVC). Ground reaction force (GRF), impulse (I), and reactive strength index (RSI) were assessed with a force platform. Data were analyzed using a two way ANOVA. Analysis of GRF showed no significant main effects for RVC condition (p = 0.46) and no interaction for RVC condition and gender (p = 0.11). Analysis of I showed no significant main effects for RVC condition (p = 0.99) and no interaction for RVC condition and gender (p = 0.61). Analysis of RSI showed no significant main effects for RVC condition (p = 0.78) and no interaction for RVC condition and gender (p = 0.20). Remote voluntary contractions appear to offer no performance benefits for exercises such as the depth jump.

KEYWORDS: Concurrent activation potentiation, plyometrics, depth jump

INTRODUCTION: The contractions of muscles remote from a prime mover have been described as remote voluntary contractions (RVC) and have been shown to increase lower body reflexes in clinical populations (Delwaide & Toulouse, 1980; Hortobagyi et al., 2003; Pereon et al., 1995).
Additionally, RVC have been recommended to improve muscle performance and this phenomenon has been described as concurrent activation potentiation (CAP) (Ebben, 2006). Researchers studying the effect of RVC determined that a combination of jaw clenching, hand gripping, and the Valsalva maneuver was more effective than jaw clenching or hand gripping alone for enhancing knee extension performance (Ebben et al., 2008a). Remote voluntary contractions have been demonstrated to be effective during isometric (Ebben et al., 2008a; Sasaki et al., 1998), but not during some dynamic tasks (Sasaki et al., 1998). On the other hand, only one published study examined the effect of RVC's during dynamic athletic movements such as jumping (Ebben et al., 2008b). In this study, subjects produced 19.5% higher rate of force development (RFD) and 20.2% faster time to peak force during the countermovement jump while jaw clenching, compared to a non-jaw clenching condition (Ebben et al., 2008b). However, these subjects did not produce greater peak ground reaction force (GRF) in the jaw clenching condition. Thus, the potential of RVC as a potentiation phenomenon for dynamic athletic tasks remains uncertain as does its potential application during slow and fast stretch shortening cycle activities.
Schmidtbleicher (1992) defined fast and slow stretch shortening cycle activities as those lasting approximately 100 ms and more than 250 ms, respectively. Countermovement jumps, which demonstrate RVC mediated improvement in performance, are thought to be slow stretch shortening cycle activities whereas the depth jump typically has been defined as a fast stretch shortening cycle activity (Schmidtbleicher, 1992). At present, it is not known if the mechanisms associated with CAP is present and potentially ergogenic during fast stretch shortening cycle activities such as the depth jump.
Therefore, the purpose of this study was to compare conditions that included RVC and a condition that did not (NO-RVC) and the effect on kinetic parameters of depth jump performance.

METHODS: Subjects included 13 men (mean ± SD, age = 21.3 ± 1.6 yr; body mass = 87.1 ± 15.7 kg; vertical jump = 62.62 ± 8.61 cm) and 10 women (mean ± SD, age 20.9 ± 1.1 yr; body mass 65.7 ± 4.35 kg; vertical jump = 45.46 ± 4.93 cm). All subjects participated in intercollegiate or recreational athletics as well as lower body resistance training and plyometrics for at least 2 months. Exclusion criteria included any history of lower limb pathology that resulted in functional limitation of the exercises to be assessed in this study. The subjects were informed of the risks associated with the study and provided informed written consent. The study was approved by the institution's internal review board.

Subjects performed a pre-test habituation and test session. Prior to each, subjects warmed up for 5 minutes with light exercise on a rowing ergometer followed by dynamic stretching. A pre-test habituation session was conducted to teach and allow the subject to correctly perform the depth jump with a subsequent vertical jump (DJ) to be used during the test session. Subjects' countermovement jump height was also assessed using a Vertec (Sports Imports, Columbus, OH, USA).

During the test session, subjects performed 2 repetitions of the DJ in the RVC and NO-RVC conditions. In the RVC condition, subjects were instructed to maximally clench their jaw on a dental vinyl mouth guard (Cramer Products Inc., Gardner, KS), clench their fists forcefully and perform a brief Valsalva maneuver during the contact phase of the DJ landing prior to the subsequent vertical jump. The NO-RVC condition included the subjects performing the DJ with an open mouth and pursed lips to limit the likelihood of jaw clenching, and cycling between inspiratory and expiratory flow in order to reduce the Valsalva effect. These methods were similar to those previously used (Ebben et al., 2008a). Depth jumps were performed from a box height that was normalized to the subjects' countermovement height assessed during the habituation session. The order of the RVC and NO-RVC conditions was counterbalanced. Five minutes of rest was provided between each condition to reduce fatigue effects. Subjects were instructed to perform maximally and were encouraged equally for all test exercises.

The test exercises were assessed with a 60 x 120 cm force platform (BP6001200, Advanced Mechanical Technologies Incorporated, Watertown, MA USA). The force platform was calibrated with known loads to the voltage recorded prior to the testing session. Kinetic data were collected at 1000 Hz, real time displayed and saved with the use of computer software (BioAnalysis 3.1, Advanced Mechanical Technologies, Incorporated, Watertown, MA USA) for later analysis. Peak vertical GRF, impulse (I), and reactive strength index (RSI) were calculated from the concentric phase of the force-time records consistent with methods previously used (Flanagan et al., 2008; Jensen & Ebben, 2007; Jensen et al., 2008). All values were determined as the average of 2 trials for each exercise. Peak GRF was defined as the highest vertical GRF value attained during the contact phase of the DJ, minus body mass (Jensen & Ebben, 2007). Impulse was calculated as the force multiplied by the time it took to develop it based on the area under the curve of the contact phase of the force time record (Jensen et al., 2008). Reactive strength index was calculated as jump height divided by the contact time (Flanagan et al., 2008).

All data were analyzed with SPSS 16.0 using a two way ANOVA to evaluate the differences between the RVC conditions and the interaction between RVC conditions and gender. The *a priori* alpha level was set at $p \leq 0.05$. The trial to trial reliability of each dependent variable was assessed using single and average measures intraclass correlation coefficients (ICC). In addition, a repeated measures ANOVA was used to confirm that there was no significant difference ($p > 0.05$) between the trials for each dependent variable.

RESULTS: Analysis of GRF showed no significant main effects for RVC condition ($p = 0.46$) and no interaction between RVC condition and gender ($p = 0.11$). Analysis of I showed no significant main effects for RVC condition ($p = 0.99$) and no interaction between RVC condition and gender ($p = 0.61$). Analysis of RSI showed no significant main effects for RVC condition ($p = 0.78$) and no interaction between RVC condition and gender ($p = 0.20$). Data are presented in Table 1. Single and average measures ICC are presented in Table 2.

Table 1. Data are presented as mean ± SD for the depth jump peak vertical ground reaction force (GRF), impulse (I), and reactive strength index (RSI) in the RVC and NO-RVC conditions.

	Men (N=13)			Women (N=10)		
	RVC	NO-RVC	%	RVC	NO-RVC	%
GRF (N)	2303.29±436.88	2495.97±585.44	7.7	1706.72±339.13	1634.47±204.82	4.2
I (N·sec)	729.19±135.74	735.71±144.38	1.0	467.68±86.50	461.03±69.38	1.3
RSI	0.83±0.23	0.82±0.30	1.3	0.60±0.19	0.57±0.13	5.0

% = percent difference between RVC and NO-RVC condition

Table 2. Intraclass correlation coefficients (ICC) for the depth jump peak vertical ground reaction force (GRF), impulse (I), and reactive strength index (RSI) in the RVC and NO-RVC conditions.

	Single Measures ICC	Average Measures ICC
RVC depth jump GRF	0.82	0.90
NO-RVC depth jump GRF	0.92	0.96
RVC depth jump I	0.87	0.93
NO-RVC depth jump I	0.93	0.96
RVC depth jump RSI	0.90	0.95
NO-RVC depth jump RSI	0.94	0.97

DISCUSSION: This is the first study to investigate the effects of CAP during a fast stretch shortening cycle activity such as the DJ. Results demonstrate that no ergogenic advantage was accrued for any of the outcome variables assessed in the RVC compared to the NO-RVC condition. Furthermore, no gender differences were found for any of the variables.

The results of this study stand in contrast to previous research that demonstrated 15.8% increases in the rate of force development during an isometric hand grip task (Hiroshi, 2003) and 14.8% increase in isometric knee extensor torque (Ebben et al., 2008a), while using RVC. Additionally, the findings of the present study are dissimilar to those that demonstrated 19.5% higher rate of force development during the countermovement jump (Ebben et al., 2008b). Thus, RVC appear to offer an ergogenic advantage during countermovement jumps, but not depth jumps. This finding indicates that RVC may work for slow but not fast stretch shortening cycle activities as defined by Schmidtbleicher (1992).

The stretch shortening cycle is known to include both passive mechanical and active neurophysiological force producing components. Previous research has demonstrated that DJ produce lower levels of muscle activation compared to many other plyometric exercise variations including the countermovement jump (Ebben et al., 2008c). This finding is thought to be due to a disproportionate reliance on passive, and not active, force producing mechanisms during the DJ (Ebben et al., 2008c). Previous reports have indicated that RVC may function due to motor overflow and a concomitant increase in muscle activation of the prime mover (Ebben 2006). Thus, fast stretch shortening cycle activities such as the DJ may occur too quickly to take advantage of RVC mediated increases in muscle activation. Based on the results of this study, this effect appears to be similar for men and women.

CONCLUSION: Results of this study demonstrate that no performance augmentation is experienced during the RVC compared to the NO-RVC condition during DJ. Remote voluntary contractions may not work during fast stretch shortening cycle activities that may preferentially rely on passive force production more than active force producing phenomena such as muscle activation.

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KINETIC ANALYSIS OF LOWER BODY RESISTANCE TRAINING EXERCISES

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This study evaluated and compared the peak vertical ground reaction force (GRF) and rate of force development (RFD) for the eccentric and concentric phases of 4 lower body resistance training exercises, including the back squat, deadlift, step-up, and forward lunge. Sixteen women performed 2 repetitions of each of the 4 exercises at a 6 repetition maximum load. Kinetic data were acquired using a force platform. A repeated measures ANOVA was used to evaluate the differences in GRF between the exercises. Results revealed significant main effects for GRF both the eccentric ($p \leq 0.001$) and concentric ($p \leq 0.001$) phases. Significant main effects were also found for RFD for the eccentric ($p \leq 0.001$) and concentric phases ($p \leq 0.001$). Force and power requirements and osteogenic potential differ between these resistance training exercises.

KEYWORDS: resistance training, ground reaction force, rate of force development

INTRODUCTION: Quantification of the intensity of training stimuli enables practitioners to select optimal exercises to elicit adaptations based on individual needs. The magnitude of muscle activation and the amount and rate of force development are of particular interest because these variables provide insight into the physical demands of resistance training exercises. Surface electromyography (EMG) and force platforms are two frequently utilized instruments that measure these variables of lower body resistance training exercises.

Surface electromyography has been used to evaluate single lower body resistance training exercises and variations therein (Ebben & Jensen, 2002; Schwanbeck et al., 2009), as well as multiple lower body exercises (Ekstrom et. al, 2007; Ebben, 2009; Ebben et al., 2009). While EMG is a valid and reliable tool for quantifying muscle activity, the amplitude of the EMG signal cannot be assumed to be equal to force production of the muscle due to several physiologic and technical factors (Neumann, 2010). However, other instruments, such as a force platform, are able to quantify kinetic variables.

Kinetic data demonstrate the magnitude of forces applied and received by the body and how quickly these forces are generated. The magnitude and rate of force generation are components of power production, which is a key determinant of athletic success for many sports (Stone, 1993). Additionally, the magnitude and rate of loading of the axial skeleton are essential determinants of the osteogenic potential of an exercise (Skerry, 1997). Exercises that promote osteogenesis are of particular importance to female athletes, who are at increased risk of impaired bone health associated with prolonged periods of amenorrhea, compared to male and eumenorrheic female counterparts (Jurimae & Jurimae, 2008).

Previously, kinetic analysis has been used to assess variations of a single exercise (Wallace et al., 2006; Wilson et al., 2008), and multiple modes of exercises, such as resistance training, plyometrics, and aerobic exercise (Ebben et al., 2009b; Morrissey et al., 1998). However, no previous study has performed a kinetic analysis of multiple variations of a single exercise mode, such as resistance training. Therefore, the purpose of the present study was to measure and compare the ground reaction force (GRF) and rate of force development (RFD) for both the eccentric and concentric phases of 4 resistance training exercises, including the back squat, deadlift, step-up, and forward lunge.

METHODS: Subjects included 16 university women whose descriptive statistics are presented in Table 1. Inclusion criteria consisted of women subjects who were 18-27 years old and were either NCAA Division I or club sports athletes, or recreationally fit, and participated in lower body resistance training for at least two days a week for at least 6 weeks. Exclusion criteria included any orthopedic lower limb pathology that restricted athletic functioning, known cardiovascular pathology, and inability to perform exercises with maximal effort. All subjects provided informed consent prior to the study, and the university's internal review board approved the study.

Table 1. Descriptive data (mean ± SD)

	Subjects (N=16)
Age (years)	21.19 ± 2.17
Height (cm)	169.39 ± 7.54
Body weight (kg)	66.08 ± 9.91
High school sports participation (years)	3.91 ± 0.38
College sports participation (years)	1.25 ± 1.91
Plyometric training participation (days/week)	0.94 ± 1.23
Resistance training participation (days/week)	2.72 ± 0.48
Aerobic training participation (days/week)	4.09 ± 2.08

Subjects attended two sessions, including one pre-test habituation session and one testing session. At the beginning of the each session, subjects participated in a standardized general and dynamic warm-up. During the pre-test habituation session, subjects were familiarized with and performed each of the 4 test exercises, including the back squat, deadlift, step-up using a 45.72 cm box, and forward lunge, in order to determine their 6 repetition maximum (RM).
Approximately 1 week after the pre-test habituation session, subjects returned for the testing session. Subjects performed 2 full range of motion repetitions using their previously determined 6 RM loads, for each of the test exercises. Randomization of the exercises, limited repetitions, and 5 minutes of recovery were provided between test exercise in order to reduce order and fatigue effects. All exercises were performed according to the methods previously described (Earle & Baechle, 2000) with the exception that the step-up began on top of the box so that all exercises consistently started with the eccentric phase and ended with the concentric phase.
All exercises were performed on a force platform (Advanced Mechanical Technologies Incorporated, Model BP6001200) that was mounted flush with a weightlifting platform to minimize risk of injury. Kinetic data were analyzed for GRF and RFD for both the eccentric and concentric phases of each of the 4 exercises. Rate of force development was calculated as the difference between the peak GRF and the GRF from a point 100 ms before the peak, divided by 100 ms. All values were averaged using the 2 test trials.
Data were evaluated with SPSS 16.0 for Windows (Microsoft Corporation, Redmond, WA, USA) using a repeated measures ANOVA to determine statistical differences in kinetic data between the exercises. Significant main effects were further evaluated using Bonferroni adjusted pairwise comparisons. Assumptions for linearity of statistics were tested and met. Statistical power (d) and effect size (η^2) are reported, and all data are expressed as means ± SD.

RESULTS: Analysis of GRF showed significant main effects for both the eccentric ($p \leq 0.001$, $d = 1.00$, $\eta^2 = 0.838$) and concentric ($p \leq 0.001$, $d = 1.00$, $\eta^2 = 0.479$) phases, indicating differences in force requirements between the exercises. Significant main effects were also found for the RFD data for both the eccentric ($p \leq 0.001$, $d = 1.00$, $\eta^2 = 0.426$) and concentric ($p \leq 0.001$, $d = 1.00$, $\eta^2 = 0.391$) phases, indicating differences in power production among the exercises. Post hoc analysis identified the specific differences between the exercises as assessed by GRF and FRD data (Table 2).

Table 2. Kinetic data (mean ± SD)

	Squat	Deadlift	Step-Up	Lunge
Eccentric GRF (N)	1473.70 ± 293.61^{c***,d**}	1416.48 ± 253.64^{c***,d**}	944.59 ± 165.74^{a***,b***,d***}	1322.50 ± 195.12^{a**,b*,c***}
Eccentric RFD ($N \cdot sec^{-1}$)	953.78 ± 759.29^{d*}	793.16 ± 570.20^{d**}	574.36 ± 252.85^{d**}	1909.65 ± 1124.69^{a*,b**,c**}
Concentric GRF (N)	1580.73 ± 361.08^{c**,d**}	1520.48 ± 276.45^{c**,d**}	1202.39 ± 216.38^{a**,b**,d*}	1399.00 ± 207.33^{a**,b**,c*}
Concentric RFD ($N \cdot sec^{-1}$)	714.86 ± 630.76^{c**,d**}	893.50 ± 456.63^{c*,d**}	2496.00 ± 1802.58^{a**,b*}	1441.69 ± 509.71^{a**,b**}

GRF = ground reaction force; RFD = rate of force development; a = significantly different from squat; b = significantly different from deadlift; c = significantly different from step-up; d = significantly different from lunge; * = $p \leq 0.05$; ** = p 0.01; *** = $p \leq 0.001$

DISCUSSION: This is the first known study to assess the GRF and RFD of several lower body resistance training exercises. Significant differences in GRF and RFD were found among the squat, deadlift, step-up, and lunge. The present study revealed differences in the force demands for both eccentric and concentric phases of the exercises, as assessed by GRF. Specifically, GRF data were greatest for the squat and deadlift, followed by the lunge, and the step-up. Previous research evaluating kinetic data during maximal isometric squats found peak GRF values of 2186.95 ± 377.34 N and RFD values of 2689.32 ± 804.80 N/s, which were higher than the values obtained in the current study (McBride et. al, 2006). This may be attributed to differences in the relative intensity of the squat between the two studies. Specifically, the previous study evaluated the squat under a maximal load, while the present study used a 6 RM load. Additionally, the RFD of the eccentric phase of the lunge was significantly greater than that of all the other exercises. This latter finding is somewhat consistent with previous research demonstrating that plyometric exercises, such as the depth jump, and loaded jumps such as the squat jump, yield greater RFD data than the squat (Ebben et al., 2010). This finding is potentially due to the eccentric or weight acceptance phase of the lunge, which is characterized by a rapid loading in the transition from non-weight bearing to weight bearing on the lead leg as the subject lunges forward. The RFD during the concentric phase of the step-up was significantly greater than that of the squat and deadlift, and trended to be greater than that of the lunge. Thus, the step-up and lunge provide a greater RFD stimulus than the squat and deadlift. The large force demands of the squat and deadlift may provide a more intense training stimulus in terms of GRF, though athletic power may be augmented by training with the lunge and step-up, due to the greater RFD component of these exercises during the eccentric and concentric phases, respectively.

Each of these exercises may have value as an osteogenic stimulus either through relatively high GRF or relatively high RFD, which may approximate the magnitude and rate of overload which are believed to be important osteogenic stimuli (Skerry, 1997).

Previous research examining similar resistance training exercises has also shown differences in muscle activation between the 4 exercises assessed (Ebben et al., 2009). This electromyographic data along with the kinetic data from the present study enhances the understanding of the characteristics of these exercises.

CONCLUSION: Of the 4 exercises assessed, the squat and deadlift yielded the greatest GRF, while the lunge and step-up had the greatest RFD demands. Training with a combination of these exercises may be ideal for obtaining adaptations along the force velocity continuum and for promoting osteogenesis.

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BIOMECHNICAL STRATEGY DURING PLYOMETRIC BARRIER JUMP-INFLUENCE OF DROP-JUMP HEIGHTS ON JOINT STIFFNESS

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The purpose of this study was to explore the joint stiffness of lower-extremity during plyometric barrier jump. Fourteen power-oriented track and field men of collegiate and national level volunteered to participate in the study. All performed 3 maximal effort drop jumps where they landed and immediately jumped over a 60 cm barrier after dropping from 30, 60 and 90 cm. The results showed both knee and ankle joint stiffness became progressively and significantly lower with the increment of drop heights. Modulating knee and ankle joint stiffness, mainly by the joint angles during touchdown, is the biomechanical strategy to accommodate for changes in different drop heights. Our findings suggest the increment of drop heights during plyometric barrier jump diminished the benefit from stretch-shortening cycle.

KEYWORDS: plyometric, stiffness, drop jump.

INTRODUCTION: Plyometric exercises are frequently used neuromuscular training in athletics (Meyer et al., 2005). Drop jump involved the stretch-shortening cycle (SSC) in the ankle, knee and hip muscles. Such rapid stretch supplied the elastic energy stored in muscle-tendon unit complexes during the braking phase of a SSC and elicited the stretch reflex for the greater power output during the push-off phase. Greater leg stiffness allows greater storage and release of elastic energy to increase the force of motion (Gollhofer et al., 1992; Komi, 1992; Wang, 2008). Commonly, athletes perform the drop jump at increased heights for a greater training stimulus. The purpose of this study was to determine the changes in the joint stiffness of lower-extremity associated with drop height increments when jumping over a barrier.

METHOD: Fourteen power-oriented track and field men of collegiate and national level (age: 22.5±3.5 years; body height: 177.1±5.6 cm; body weight: 87.2±16.5 kg) volunteered to participate in the study. All volunteers were enrolled after providing written informed consent. Prior to experiment, subjects changed specific footwear (New Balance Running Shoe, Model 629; New Balance Athletic Shoe, Inc., Boston, MA, USA) to control for different shoe-sole absorption properties. Lower-body lunging and squatting movements were performed as the warm-up exercise. Then they were asked to perform 3 maximal effort drop jumps where they landed and immediately jumped over a 60 cm barrier after dropping from 30, 60 and 90 cm (DJ30, DJ60, DJ90) (Figure 1). Kinematic data were collected at 240 Hz using 6 Eagle cameras which were positioned around the performance area and synchronized to force platforms (Bertec 4060 NC; Bertec Corp., Columbus, OH, USA) collected at 1200Hz. One platform recorded right extremity data, and one recorded left extremity data. The cameras and subsequent performance area were calibrated, yielding mean residual errors of 1.1-1.53 mm over a volume of 2.5 x 2.1 x 2.5 m. The marker coordinate data were processed using Orthotrak (Motion Analysis Corporation, Santa Rosa, CA, USA) and custom Matlab programs (Mathworks Inc., Natick, MA, USA). Based on a frequency content analysis of the digitized coordinate data, marker trajectories were filtered at 10.5 Hz using a fourth order Butterworth filter. Raw ground reaction force data were exported, and data were normalized to body weight. The onset of the ground contact phase was determined when the vertical ground reaction force (VGRF) exceeded 30N threshold. Net muscle joint moments (M) were calculated by combining the kinematic and force plate data with anthropometric data using the inverse dynamics solution. Positive net muscle joint moment was defined as extensor activity, while negative net muscle joint moment indicated the activity of the flexors. The joint stiffness (k_{joint}) was calculated by the formula: (M_{joint})/ ($\Delta\theta_{joint}$), where the M_{joint} was the net muscle joint moment while the joint was maximally flexed during the ground contact phase, and the $\Delta\theta_{joint}$ was the angle change of the joint in the sagittal plane between the start of the ground contact phase and the instant when the joint was maximal flexed. The normalized joint stiffness was defined as the normalized net muscle joint moment divided by the joint angle change, and the unit was $Nm \cdot kg^{-1}/deg$. A repeated measures analysis of variance (3 drop height and 2 legs) was performed on normalized knee and ankle stiffness, normalized knee and ankle moment, knee and ankle joint angle changes, knee and ankle joint angles at touchdown, and maximal knee and ankle joint angles during ground contact phase using SPSS for Windows (Version 11.0, Chicago, IL) with alpha level of 0.05.

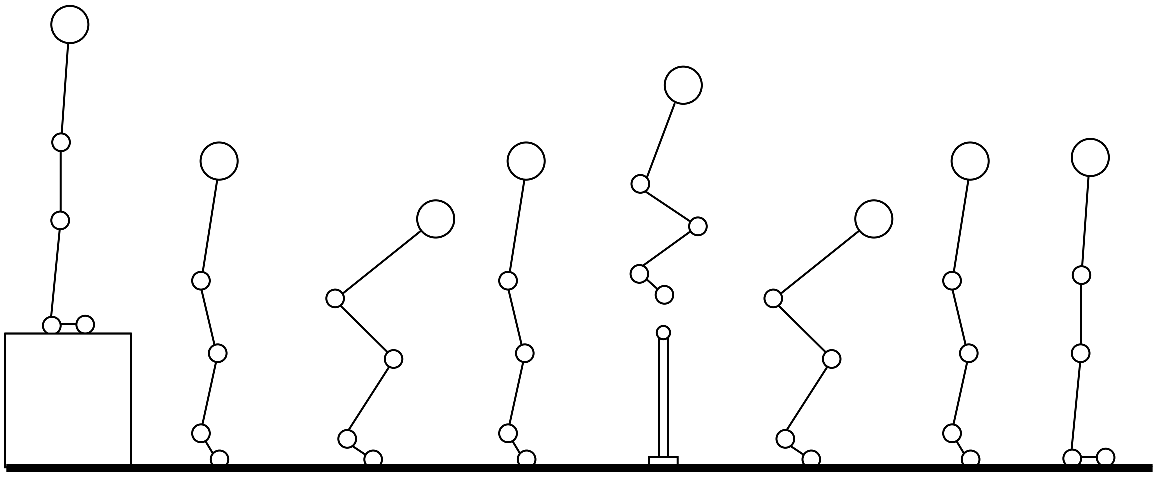

Figure 1. Modelled Representation of the Human Body during Plyometric Barrier Jump

RESULTS: There was no interaction between drop heights and legs for all outcome variables tested except for the knee flexion angle at touchdown. The normalized joint stiffness related parameters were summarized in Table 1. Both knee and ankle joint stiffness became progressively and significantly lower with the increment of drop heights (all p=0.001 and p<0.005 for pairwise comparison between each drop height for knee and ankle joint, respectively). The normalized muscle joint moment of knee joint was significantly higher in 60DJ and 90DJ than that in 30DJ (p=0.009 and p<0.005, respectively). Similarly, the ankle muscle joint moment was significantly higher in 60DJ and 90DJ than that in 30DJ (p=0.008 and p=0.022, respectively). There were no significant differences between 60DJ and 90 DJ for knee and ankle muscle joint moments (p=0.055 and p=0.451, respectively). With the increment of drop heights, both knee and ankle joint showed progressively and significantly larger angle changes during ground contact phase (all p<0.005 for pairwise comparison).

Table 1. Comparison between Drop Height Changes of Normalized Joint Stiffness Related Parameters during Plyometric Barrier Jump[a]

	30DJ		60DJ		90DJ	
	R't Leg	**L't Leg**	**R't Leg**	**L't Leg**	**R't Leg**	**L't Leg**
Knee						
Normalized M_{knee} ($Nm \cdot kg^{-1}$)	3.38(0.73)	3.49(1.23)	3.66(0.71)	3.65(1.30)	3.79(0.76)	3.76(1.34)
$\Delta\theta_{knee}$ (degree)	31.61(8.87)	31.50(9.01)	38.29(8.35)	40.57(11.21)	46.53(7.86)	47.49(9.53)
Normalized k_{knee} ($Nm \cdot kg^{-1}$/degree)	0.12(0.06)	0.13(0.06)	0.11(0.04)	0.10(0.04)	0.09(0.03)	0.08(0.03)

Ankle						
Normalized M_{ankle} (Nm·kg^{-1})	2.52(0.51)	2.66(0.71)	2.73(0.50)	2.91(0.82)	2.73(0.32)	2.81(0.82)
$\Delta\theta_{ankle}$ **(degree)**	30.77(10.18)	29.42(8.25)	43.34(7.27)	42.52(6.14)	51.50(6.81)	50.06(5.82)
Normalized k_{ankle} (Nm·kg^{-1}/degree)	0.09(0.02)	0.10(0.02)	0.06(0.01)	0.07(0.02)	0.05(0.01)	0.06(0.02)

[a] Data are presented as mean (SD)

Plyometric barrier jump with different drop heights demonstrated different landing patterns (strategies) (Table 2.). The knee and ankle joint became more straight and plantarflexed, respectively with the increment of drop heights (all p<0.005 for pairwise comparison between each drop height for both knee and ankle joint). However, they soon reached comparable degrees of maximal knee flexion and ankle dorsiflexion angles before taking-off (all p>0.05, except for maximal ankle dorsiflexion between 60DJ and 90DJ).

Table 2. Changes of Knee and Ankle Joint Angles during Ground Contact Phase of Plyometric Barrier Jump[a]

	30DJ		60DJ		90DJ	
	R't Leg	**L't Leg**	**R't Leg**	**L't Leg**	**R't Leg**	**L't Leg**
Knee						
Angle at touchdown (degree)	39.65(9.44)	40.21(9.25)	30.76(7.85)	28.15(6.14)	24.34(6.72)	23.05(5.69)
Maximal flexion angle (degree)	71.27(5.44)	71.70(6.92)	69.05(5.99)	68.72(8.32)	70.87(4.51)	70.54(6.83)
Ankle						
Angle at touchdown (degree)	0.50(9.60)	1.09(9.25)	-12.69(7.52)	-13.40(8.73)	-18.91(6.42)	-19.43(6.82)
Maximal flexion angle (degree)	31.27(4.82)	30.52(5.00)	30.65(5.25)	29.13(5.29)	32.59(4.59)	30.63(6.03)

[a] Data are presented as mean (SD)

DISCUSSION: Plyometric jumping exercises have became more and more popular in athletic training field, however, little is known about lower-extremity joint stiffness characteristics during plyometric barrier jump. Drop jumps involve a spring-like manner, where the leg spring compress and then lengthens during the ground contact phase (Farley and Morgenroth, 1999). The current study showed high-level athletes demonstrated significant decreases of knee and ankle joint stiffness while the drop heights increased from 30 to 90 cm. Comparing 60DJ with 30 DJ, both muscle joint moments and joint angle changes increased at knee and ankle joint. As the drop height increased to 90 cm, only knee flexed and ankle dorsiflexed angles increased around 8-degree without changes of lower-extremity muscle joint moment were evident. Interestingly, adjustment of knee and ankle joint stiffness, which was mainly modulating the knee and ankle angles during touchdown, was the biomechanical strategy to accommodate for changes in different drop heights. There were no differences between maximal flexion angles during ground contact phase regardless the joint. It may probably because these 3 drop jumps have the fixed barrier height in common that affect the activation of the neuromuscular system to the degree leading to achieve the success of the following barrier jump.

Joint stiffness depends on many factors, including the stiffness of each muscle-tendon unit that cross the joint (Farley and Morgenroth, 1999). Tendon and muscle stiffness increased with the force and activation level of the muscle. In the present study, the net muscle moment at knee and ankle joints increased from 30 to 60, and to 90 cm, however, these increases did not result in simultaneously increases in the knee and ankle joint stiffness. According to our previous report (Kernozek et al., 2007), the power absorption was increased with the increments of the drop height during plyometric barrier jump. The increase of knee extensor and ankle plantarflexor moments were used in the power absorption, therefore, decreased the joint stiffness. Although drop jump at increased heights could obtain a greater training stimulus, the diminishing leg stiffness during landing motion might decrease the benefit from the SSC, since greater leg stiffness allows greater storage and release of elastic energy to increase the force of motion (Gollhofer et al., 1992; Komi, 1992; Wang, 2008). Additionally, since enhanced joint angular stiffness could resist sudden angular displacement, which is beneficial to the joint stability. Previous study suggested that reduced joint angular stiffness may increase the damage to cartilage and ligaments (Butler et al., 2003).
There were some limitations of this study. The results of the current experiment were not comprehensive for every lower-extremity joint performance. Only men of advance athletic ability were used, hence the results might not relate directly to all athletic populations.

CONCLUSION: Decreased knee and ankle joint stiffness associated with the increment of drop heights during plyometric barrier jump diminished the benefit from stretch-shortening cycle. The result may serve as training basis for plyometric exercise practitioners.

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WHAT HAVE WE LEARNED FROM TEACHING CONFERENCES AND RESEARCH ON LEARNING IN BIOMECHANICS?

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A narrative review was conducted of biomechanics teaching/learning papers published in teaching conference proceedings and in journals since 1980. The majority of the papers published focused on course concepts and technology, rather than reporting data on student learning. Recent progress has been made in standardized tests of biomechanical concepts and identifying factors that are associated with learning these concepts. Future research should use these tests to focus on learning-related factors and active learning strategies from physics education research to improve student mastery of biomechanical concepts.

KEYWORDS: Instruction, kinesiology, pedagogy, physics, teaching.

INTRODUCTION: Teaching introductory biomechanics to most exercise science/kinesiology majors is a challenging task. Teaching biomechanics is difficult because the field integrates two difficult bodies of knowledge: the complexity of human anatomy with the mechanics of the body and the external forces it encounters. Add to this problem of non-linear, complex biomechanical systems, the fact that many kinesiology majors are not adequately prepared or interested in natural sciences like mechanics, and it is obvious why teaching introductory biomechanics is a challenging task for many faculty.

Biomechanics scholars have a long tradition of sharing teaching methods and materials to address this challenge. In fact, biomechanics faculty have organized five teaching conferences to discuss these issues since 1977 in North America. The purpose of this paper was to review the proceedings of these teaching conferences and the literature on teaching biomechanics to summarize the scholarship of teaching and learning (Boyer, 1990) in the discipline.

METHOD: The author reviewed the five published proceedings of the North American teaching conferences in biomechanics, as well as papers published on biomechanics of teaching and learning in journals since 1980. Papers in the five teaching proceedings (Dillman and Sears, 1978; Shapiro and Marrett, 1984; Wilkerson et al. 1991, 1997; Blackwell and Knudson, 2001), excluding summaries of discussion sessions, were classified into one of five categories based on the objectives and data in the papers: CCH-Course concepts and history, AOL-Activity or laboratory learning activity, TIP-Teaching idea or pedagogy, ETS-Equipment, technology or software, or STL-Scholarship of teaching and learning

Papers were considered STL if they reported data on student perceptions or learning outcomes from the instructional activities discussed. A review was also performed on papers identified by a search of several bibliographic databases for STL research in biomechanics. One hundred sixty-two teaching conference papers and twenty-one journal articles were reviewed.

RESULTS AND DISCUSSION: Forty-six papers were published in the proceedings of the first “national” conference on teaching “kinesiology” (Dillman & Sears, 1978). Two historical issues are important to note about this conference. First, the national conference, and the conferences that followed benefited from international participation (e.g. Canada, UK, Australia). Second, although many courses were still called “kinesiology” in 1977, the topic was really biomechanics and many of the papers focused on course content, technology, and teaching ideas. A young field was coming to grips with a lack of consistency in terminology and course content. Issues of contention were the balance of anatomy and mechanics content, qualitative and quantitative

analysis of human movement, as well as theory versus application. The several versions of the National Association for Sport and Physical Education (NASPE) Guidelines and Standards for Undergraduate Biomechanics were born from the discussions at the teaching conferences (Kinesiology Academy, 1980,1992; NASPE, 2003). The first two of three national surveys on instruction in biomechanics were presented at these conferences (Deutsch et al. 1978; Marett et al. 1984). Satern (1999) reported the most recent biomechanics teaching survey.
Subsequent teaching conferences had progressively fewer papers (40-35-24-17), even though the number of biomechanics programs and faculty expanded. While these teaching meetings have fostered collaboration and some consistency in this important core course. Relatively few papers published in the proceedings (Table 1) have presented actual data on student learning in biomechanics (STL), with the highest percentage of papers introducing technology (ETS) or sharing general concepts about teaching biomechanics (CCH).

Table 1. Kinds of Papers Presented at the National Conferences on Teaching Biomechanics

(Percentage)

Meeting	CCH	AOL	TIP	ETS	STL
1st	28	15	22	35	0
2nd	28	25	10	15	0
3rd	26	17	14	34	9
4th	21	25	8	42	1
5th	29	18	6	29	18

*CCH: Course Concepts/History, AOL: Activity or Lab, TIP: Teaching Idea/Pedagogy, EST: Equipment/Technology/Software, or STL: Scholarship of Teaching and Learning. STL involves the collection and reporting of student perceptions, interaction, or learning data that is peer-reviewed and shared with an external audience (Boyer, 1990).

Some of the earliest STL studies reporting learning data in biomechanics were reported at the 3rd national conference on teaching biomechanics. Knudson et al. (1991) used a pre- post-test design and found that traditional instruction in biomechanics did not automatically transfer to ability to qualitative analysis of sports skills, while Dedeyn (1991) reported retrospective data showing biomechanists with significantly better that visual movement analysis ability than teachers or undergraduate students. Other teaching conference STL papers have also utilized pre- and post-test measures of student learning. McPherson and Guthrie (1991) examined the addition of computer-assisted instruction (CAI) in introductory biomechanics. There was no significant effect of the additional CAI, even though the student attitudes about CAI were positive. Bird et al. (1997) reported preliminary data that instruction could improve mastery of NASPE standards from 18 to 74%. McGee, Fletcher and Bird (1997) used similar methodology and reported a 10% increase in learning of EMG concepts following hands-on EMG experiences compared to classroom instruction. Coleman (2001) integrated a standardized physics test (Force Concept Inventory) into the introductory biomechanics course at the University of Edinburgh and found that mastery of Newton’s Laws of motion improved from about 30% to 70%.

Knudson and colleagues (2003) implemented Coleman's suggestion to create a standardized test of biomechanical concepts. The Biomechanics Concept Inventory (BCI) is a 24 question test based on the NASPE standards (NASPE, 2003) validated with a national sample of classes. Research using the BCI test and subsequent versions has been remarkably consistent with the physics education research (PER) that improvement in mechanics knowledge falls short of instructor objectives. Typically, mean improvement is between 25 and 40%, which is equivalent to about 20% of individual maximum possible improvement. Dixon (2004) also reported a pre- post-test for biomechanics instruction in exercise science.
Research using the BCI has identified variables that are associated and are not associated with student learning of biomechanical concepts. Course and instructor characteristics account for much smaller variance (2-5%) in learning (Knudson et al. 2009) than student characteristics and behaviors (14 – 40%) do (Hsieh & Knudson, 2008; Hsieh et al. 2010). Another important observation was that increasing course credit hours from 3 to 4 with a laboratory (66% increase in contact hours) significantly improved (doubled) learning (Knudson et al. 2009). Knudson et al. (2009) also reported a weak inverse association (r = -0.18) between average spending on labs and learning. This was interpreted as a possible distraction effect of technology that has also been reported in hypermedia and visualization research (Chandler, 2009), and was also a caution about "black box" use of computers in biomechanics instruction noted by Miller (1997). Student learning of biomechanical concepts is primarily related to grade point average, and student's perception of career relevance, and their interest in the subject.
The papers on teaching biomechanics published in journals mirror the distribution of papers from the teaching conferences. Most papers have also focused on proposed applications of instructional technology (Carlton et al. 1999; Chow et al. 2000; Kirtley and Smith, 2001; Nicol and Liebscher, 1983) or lab activities (e.g. DiCarlo et al. 1998). There is little STL research on instruction using these technologies or other teaching methods to determine if these new tools and ideas increase learning beyond traditional instruction with biomechanics students.
Several biomechanics STL studies have been reported the biomedical engineering literature, focusing on using computer-assisted, active-learning or challenge-based instruction. These studies are based on several decades of PER reporting significant improvements in learning mechanical concepts with these pedagogies (Hake, 1998). Three studies reported no significant differences with these instructional innovations (Duncan and Lyons, 2008; Roselli and Brophy, 2006; Washington et al. 1999), while Pandy et al. (2004) reported significant improvements in learning with challenge-based instruction compared to traditional instruction for students within a class. These results highlight the difficulties in creating new active learning pedagogies in biomechanics. Good summaries of these active learning strategies in PER have been reported (Hake, 1998; Henderson and Dancy, 2009; Redish and Steinberg 1999) and the PER Central web site provides electronic access to some journals and research in this area (http://www.compadre.org/per/index.cfm).

CONCLUSION: Only a small percentage of papers from previous teaching conferences and journals report data-based research on student learning in biomechanics. Future research should focus on measures of student learning of biomechanics concepts and explore active learning strategies that have been effective in PER. Biomechanics-specific research supports the hypothesis that for new technologies or pedagogies be effective, they must be designed to accommodate student's abilities and be attentive to student attitudes toward biomechanics.

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MOVEMENT ANALYSIS FOR JAVELIN THROWERS IN THE QATAR 2009 CHAMPIONSHIPS

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A biomechanical analysis of the javelin throw at the Qatar Athletics Championships in Doha with comparison to international throwers was carried out by the Qatar Olympic Committee in 2009. This paper presents the results of this study for male finalists. The methodology used is based on Video Photogrammetric analyses at 50Hz. The results show the characteristics of the throwers' individual model at the event, which for practical purposes can be compared with the performance of the same throwers in other competitions. In general the Qatari throwers held the javelin longer and threw at a lower. Variations in throwers' technique may result in differences in performance.

KEY WORDS: Javelin, dynamics, force,

INTRODUCTION: A description of the technique used by elite throwers gives insight into individual forms to obtain high performance. These models become references that help coaches and athletes to develop their own strategies to achieve maximum efficiency.
The pattern of motion used in the javelin throw is similar to other movements used when striking or throwing an object (Atwater, 1979; Menzel, 1987). These are characterized by the fact that the body segments act sequentially to attain the maximum speed in the most distal segment of the system at the instant when the object is struck or thrown. The present paper describes the technical models used by two Qatari finalists in the men's javelin competition at the Qatar Athletics Championships in Doha in March 2009. The aim of the study was to compare the throwers' individual models in the light of the documented data available on the biomechanical analysis of javelin throw.

METHODS: All throws in the final were filmed and the best attempts of both athletes were subsequently analyzed. The camera (SVHS Panasonic video cameras, operating at 50 fps) was aligned with the optical axis at approximately 90 degree angle to the side view. A modulated reference system (one integrated square of 2x2 m) was used for spatial calibration. The throws were analyzed using Dartfish v.5 software. The selected timing and other kinematic parameters were obtained from the digitized coordinates.
The biomechanical analysis of each athlete focused on the Preparatory and Final Delivery phases. The most important factors for javelin release occur during these decisive periods, which therefore offer the best comparison for athletes' techniques (Campos et al., 2000).
The main time points were the following:

- t1: right foot lands (support leg for right-handed thrower) on the ground (single-support) at the beginning of the Preparatory Phase
- t2: left foot lands (braking leg for right-handed throwers) on the ground (double-support or Power Position) at the end of the Preparatory Phase and at the beginning of the Final Delivery Phase
- t3: javelin is released (instant of release) at the end of the Final Delivery Phase.

The reference values from three international elite athletes were obtained, for comparison, from biomechanical analyses at the 12th World Championships in Athletics in 2009 (IAAF, 2009).

RESULTS AND DISCUSSION: ***Duration of the Preparatory Phase and Final Delivery Phase:*** The results show that the greatest differences between the athletes occur in the Preparatory Phase. Times recorded for the duration of the Preparatory Phase ranged from about 0.13 s for the international throwers up to 0.19 s for a Qatari thrower (Table 1). The international throwers had wider variation in the Final Delivery Phase, whilst the Qatari throwers had very similar times. As an average, the Qatari throwers had longer durations in both phases than the international throwers with 0.07 seconds more spent as total time for these phases.

Table 1. The results of all subjects and duration of the different phases

Athlete and nationality	Final result [m]	Preparatory phase [s]	Final Delivery Phase [s]	Total time [s]
Thorkildsen (NOR)	89.59	0.14	0.18	0.32
Martinez (CUB)	89.41	0.13	0.26	0.39
Murakami (JAP)	82.97	0.13	0.20	0.33
International Average		0.13	0.21	0.34
Ahmed (QAT)	76.98	0.12	0.26	0.38
Ibrahim (QAT)	67.99	0.19	0.25	0.44
Qatari Average		0.16	0.26	0.42
Difference between averages		0.03	0.04	0.07

Duration between maximum (peak) joint speed and the instant of release: The quality of energy transfer to the javelin is influenced by the coordinated motion of the upper limb, starting from the acceleration-deceleration of the sequences in the upper kinetic chain. These sequential motions from proximal to distal segments are one of the fundamental keys to performance in over-arm throwing (Atwater, 1979; Mero et al., 1994). Hip, shoulder, elbow, hand and javelin velocities were taken into account to analyse power transmission sequences in delivery.

The analysis of how the maximum speed timing for each marker are reached during delivery (table 2) provides a more detailed description of the timing used by the throwers to structure their individual motion models for the upper limb.

Table 2. Peak joint speed timings during delivery

	Total time	Hip time		Shoulder time		Elbow time	
Athlete and nationality	[s]	[s]	[%]	[s]	]%[	[s]	]%[
Thorkildsen (NOR)	0.32	0.12	38%	0.10	31%	0.06	19%
Martinez (CUB)	0.39	0.14	36%	0.08	21%	0.05	13%
Murakami (JAP)	0.33	0.12	36%	0.08	24%	0.06	18%
International Average		0.13	37%	0.09	25%	0.06	17%
Ahmed (QAT)	0.38	0.14	37%	0.08	21%	0.05	13%
Ibrahim (QAT)	0.44	0.14	32%	0.10	23%	0.06	14%
Qatari Average		0.14	35%	0.09	22%	0.06	14%
Difference between averages		0.01	-2%	0.00	-3%	0.00	-3%

Table 2 shows the data of time duration from maximum hip, shoulder and elbow speed to release, with average times of 0.13 s for the time from maximum hip speed to release, 0.09 s from maximum shoulder speed to release and 0.06 s from maximum elbow speed to release for the International throwers. The two Qatari throwers showed similar timings. The differences between the International and Qatari throwers were in relative timings. The Qatari throwers displayed maximum hip speed 2% later, and both shoulder and elbow speed 3% later than the international throwers. These differences in starting hip motion confirm findings by Best et al. (1993) that this parameter depends on individual technique and its effect on performance should be considered in relative terms.

Release conditions (release height; release angle and angle of attack): Release height is a measure of ballistic efficiency and depends on the thrower's height, lateral bending of the trunk and front leg knee angle at the instant of release (Mahmud, 2007). Throwers should aim to release the javelin from as high as their height allows while maintaining foot contact on the ground. The results show release heights that range from 1.80 m to 2.14 m. The parameters relative to the position of the javelin at release should include javelin position angle (also called attitude angle), release angle and, as a consequence of these, angle of attack. (Mahmud, 2007).Attitude angle is the angle between the position of the javelin and the horizontal, the release angle is formed by the velocity vector and the horizontal, and the angle of attack is the difference between attitude angle and release angle. Theoretical references suggest that the release angle should be 32° - 37° and the angle of attack not over + 8° for an effective throw. (Morris et al., 2001).

Knee Angle of the braking leg (Final Delivery Phase t2 - t3): The bracing and blocking action of the braking leg must also be taken into account in order to reach maximum release velocity, as it greatly reduces the horizontal velocity of the thrower-plus-javelin system (Morris et al., 2001).The knee angle of the braking leg is an indicator of the athlete's ability to transfer kinetic energy to the javelin. This blocking action favors kinetic energy transfer from the upper part of the body to the javelin. It seems evident that this action is decisive, considering that in elite throwers 60% of the javelin's kinetic energy is generated in the last 50 ms before release (Morriss & Bartlett, 1995). Results from the international throwers showed an average knee angle of 162° at t3, but the Qatari average was found to be considerably less at 151° (table 3). The Qatari throwers do not seem to be able to hold the knee angle of the braking leg as stable as the international throwers.

Table 3. Braking leg knee angle values at t1, t2& t3

Athlete and nationality	t1	t2	t3
Thorkildsen (NOR)	170	162	169
Martinez(CUB)	171	152	153
Murakami (JAP)	178	163	166
International Average	173	159	162
Ahmed(QAT)	175	141	147
Ibrahim(QAT)	173	145	155
Qatari Average	174	143	151

All the finalists showed increasing extension of the braking leg knee angle in the Final Delivery Phase. Therefore, braking leg knee extension at release was higher than for the whole of the Final Delivery Phase. The main difference between the International throwers and Qatari throwers were that the Qatari throwers knee angle was very flexed at t2, i.e. already at the instant of left foot landing. Thus, although both International and Qatari throwers extended their knee from this point onwards, due to already more extended knee at the start of the final delivery phase, the international throwers ended up having better support (equal to a more extended knee) at the time of release (t3). This allows a higher release point and higher release velocity.

Hip and Shoulder axis rotation on the sagittal plane: Rotation of the hip and shoulder axis in the sagittal plane are two important measures that show the thrower's ability to make a wide and continuous movement in the Final Delivery Phase and help throw the javelin further. With regard to shoulder motion, both International and Qatari averages were 175° at the start of the preparation phase (table 4). During this phase, the international throwers rotated the shoulder more than Qatari athletes and ended up with an angle of 133° at the start of double-support (t2), which is in line with a study by (Morris & Bartlett 1996) on elite throwers. In addition, there was greater variability in the difference between shoulder and hip axes angles at t1 than at t2. The Qatari athletes' hip angles were considerably smaller than the international athletes' hip angle at t1, but then larger at t2. Similarly to hip, the shoulder rotation undergone by the Qatari athletes was less than for the international athletes showing that the Qatari athletes should get a better body position at t1 and then have stronger rotation in the preparation phase.

Table 4. Measurements recorded for each athlete during t1 and t2.

	Hip		Shoulder		Hip-Shoulder difference	
Athlete with nation	t1	t2	t1	t2	t1	t2
Thorkildsen (NOR)	141	107	165	133	24	26
Martinez(CUB)	182	114	180	135	-2	21
Murakami (JAP)	170	114	181	132	11	18
International Average	164	111	175	133	11	21
Ahmed(QAT)	124	124	188	154	64	32
Ibrahim(QAT)	135	111	162	143	27	32
Qatari Average	129	117	175	149	46	32

CONCLUSIONS: In agreement with previous studies it was observed that each thrower maintained an individual throwing pattern in relation to timing and the values obtained in the different kinematic parameters under study. Nevertheless, these individual patterns are related to what could be called efficiency filters. These are the minimum requirements needed to throw the javelin a long distance which affect the position of the kinetic chain in the Final Delivery Phase as well as the coordination of the body segments for ballistic movement. The Qatari athletes showed weak individual patterns on the body position in the final delivery.
The aspects that distinguished Thorkildsen from the rest of the throwers was that his movements were more rectilinear in the final phases and he throws from a higher position, with a longer acceleration path and more favorable release conditions. However, It is felt that the information presented herein will be useful for javelin throw coaches and throwers and that it will contribute to the understanding of this sport and improve their achievement.

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ADJUSTMENT OF THE LOWER LIMB MOTION AT DIFFERENT IMPACT HEIGHTS IN BASEBALL BATTING

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The purpose of this study was to investigate the change in the lower limb motion to three different hitting areas of the strike zone: high, middle, and low. Subjects were 10 right-handed male skilled batters of a university baseball team. Data were collected using a three dimensional automatic motion analysis system (Vicon 612). Joint angles of the lower limbs were computed. Comparison of the hitting in the high area vs. low area revealed that to hit the ball in the low area the batter adjusted the motion of the hip joint by regulating the flexion-extension angle of the both hips from the phase of the Swing start to the phase of the Impact. After that the phase of the Left upper arm parallel abduction angle of the right hip was smaller in case of the high, middle areas than the of the low area, and abduction angle of the left hip was larger in case of the high, middle areas than the low area.

KEY WORDS: three dimensional motion analysis, angular kinematics, striking.

INTRODUCTION: Many investigations of baseball batting have analyzed the techniques by which a batter hits a ball in the middle of the hitting areas (McIntyre, 1982; Messier 1985). However, since the pitching course in actual game varies, the batter has to modify and change the batting swing so that he or she reacts to various areas. There is little information of how a batter modifies the motion to various pitching courses. Tago et al. (2006) reported that in case of the high, middle hitting areas, the rotation of the shoulder at the impact phase was larger than the low hitting areas. Tago et al. (2009) indicated that in hitting a ball in the high and low areas, the batter adjusts the position of the bat by modifying shoulder and elbow angles, particularly at Left upper arm parallel (LUP) and modifies the angles in the left upper limb just before impact. The purpose of this study was to investigate the change in the lower limb motion to the different hitting areas.

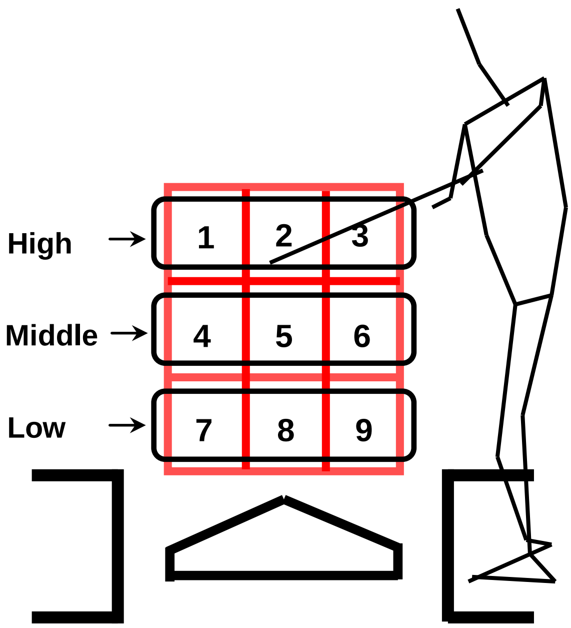


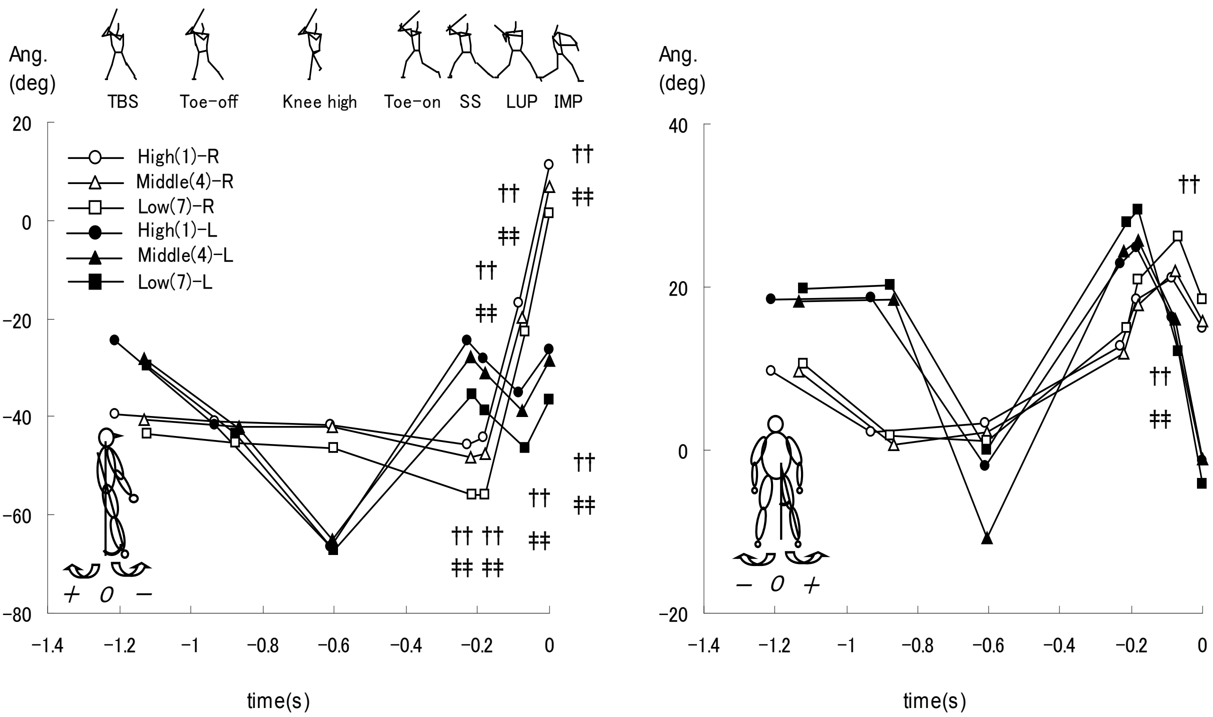
Fig.1 Hitting areas set in this study

METHODS: The experimental procedure and the setting were conditions similar to Tago et al. (2009). Subjects were ten right-handed batters of a university baseball team. Informed consent was collected after the explanation of the experimental procedure. Three different hitting areas were set in accordance to the rules of baseball. The batting tee commonly used during practice was used to modify hitting areas. The high areas for right-handed batters were defined as 1, 2, and 3 of Figure1, the middle areas as 4, 5 and 6 of Figure1, the low areas as 7, 8 and 9 of Figure1. The subjects were given the hitting areas in random order, and the position of non-stride leg was set as the same position at the beginning. The coordinate axes were defined as follows: the Y axis was set as the direction to a pitcher, the X axis as the medio-lateral direction, and the Z axis as the perpendicular direction. Data was collected by using a three dimensional automatic motion analysis system (Vicon 612). Nine cameras operating at 250Hz were used to capture the players' motion. From several trials for each point, one trial of the fastest ball velocity and the best self-evaluation was chosen in each point and subject for analysis.

For the analysis and description of data, the batting swing was divided by seven instants as follows : Start of take back (TBS): The phase at which the bat grip began to move toward a catcher. Toe-off: The phase at which the stride leg broke the contact with the ground. Knee-high: The phase at which the knee of the stride leg was in the highest position. Toe-on: The phase at which the tip of the foot of the stride leg contacted with the ground. Swing start (SS): The phase at which the bat grip began to move toward a pitcher. LUP: The phase at which the left upper arm of the batter was in parallel to the X-axis (L-upper arm parallel). Impact (IMP): The phase at which the bat contacted with the ball.

Angular kinematics computed were joint angles of the right and left ankles, knees, and flexion-extension, adduction-abduction angle of the hips. Two-way ANOVA (three heights X three courses) was used to examine the difference in the angular kinematics of the phases mentioned above between hitting areas, setting significant level at p = 0.05.

RESULTS AND DISCUSSION: Figures 2 and 3 show the average joint angles at seven phases during hitting in the high and low hitting areas. Figure 2-1 shows Flexion-Extension angles of the hips and Figure 2-2 shows adduction-abduction angle of the hips. In the figures,



1. Flexion-extension angle of the hip joint

2. Adduction-abduction angle of the hip joint

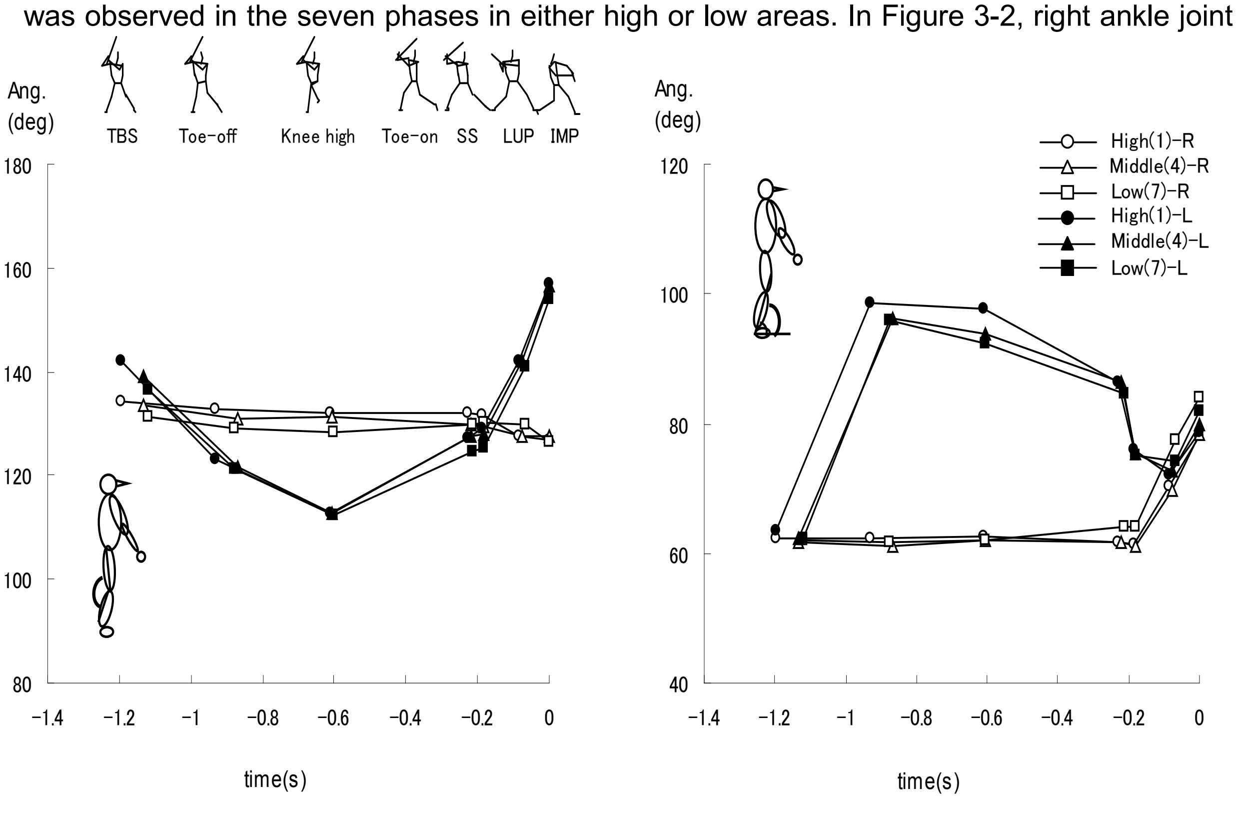
Fig.2 Changes in the angle of the hip joint during batting in height hitting areas.

Significant difference ($p<0.05$) : High vs. Low ; ††, Middle vs. Low ; ‡‡

one example is shown from the present study, R indicates the right limb, L is the left limb, and, and (1),(4),(7) indicates the hitting area (Refer to Figure.1). Significant differences are shown by a symbol († †, ‡ ‡). And the definition of the each joint angle is shown in the picture in the graph.

In Figure 2-1, flexion angle of the right hip was almost constant until the phase of the SS. After that the extension angle of the right hip quickly increased toward the impact in both high and low areas. The significant difference was observed at the SS to IMP, i.e. the flexion angle of the right hip at the low area was larger than that of the high and middle areas. Flexion angle of the left hip was quickly increased until the phase of the Knee-high. After that the extension angle of this joint suddenly increased toward the impact in both high and low areas. The significant difference was observed at the Toe-on to IMP, i.e. the flexion angle of the left hip at the low area was larger than that of the high area. In Figure 2-2, abduction angle of the right hip was gradually increased from the phase of the Toe-off to the impact in both high and low areas. The significant difference was observed at the LUP, i.e. the abduction angle of the right hip at the low area was larger than that of the high area. Adduction angle of the left hip suddenly increased toward the Knee high. After that the abduction angle of this joint suddenly increased toward the SS, and again adduction angle of this joint abruptly increased toward the impact in both high and low areas. The significant difference was observed at the LUP, i.e. the adduction angle of the left hip at the low area was larger than that of the high and middle areas.

In Figure 3-1, right knee joint angle was almost constant until the phase of the IMP. However, no significant difference was observed in the seven phases in both high and low areas. The left knee joint flexed until the phase of the Knee high. After that this joint quickly extended toward the phase of the IMP in both high and low areas. However, no significant difference was observed in the seven phases in either high or low areas. In Figure 3-2, right ankle joint



1. Knee joint angle 2. Ankle joint angle

Fig.3 Changes in the angle of the knee and ankle joint during batting in height hitting areas.

was almost constant until the phase of the SS. After that this joint suddenly extended toward the phase of the IMP in both high and low areas. However, no significant difference was observed in the seven phases in either high or low areas. Left ankle joint suddenly extended until the phase of the Toe-off. After that this joint gradually flexed toward the phase of the IMP in both high and low areas. However, no significant difference was observed in the seven phases in either high or low areas.
Comparing hitting the ball in the high area with the low area, we will be able to identify that hitting a low compared with a high ball was characterized by; the batter adjusted the motion of the hip joint by regulating the flexion angle of the both hips from the phase of the SS to the phase of the IMP. At the phase of the LUP abduction angle of the right hip was smaller in case of the high, middle areas than the case of the low area, and abduction angle of the left hip was larger in case of the high, middle areas than the case of the low area. The opposite tendency to the high area was observed in the case of the low area. The significant differences in selected joint angles were observed after the commencement of the swing, which may imply that adjustments occur during the forward swing period. It was suggested that the movement of both hip joints from which a significant difference was seen in all the joint angles of the lower limb after the commencement of the swing be especially important.

CONCLUSION: In the current study the findings imply that in hitting a ball in the high and low areas, the batter adjusts the motion of the hip joint by modifying the flexion-extension angle of both hips from the phase of the SS to phase of the IMP, and the adduction-abduction angle of the both hips at the phase of the LUP, particularly the movement of the both hips since the phase of the SS is important.

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THE 3-D KINEMATIC ANALYSIS OF DIFFERENT TENNIS SERVE

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KEYWORDS: Tennis serve, foot-up serve, foot-back serve, Kinematics

INTRODUCTION: The key to success for a good tennis player is to be able to take the advantages of serving and keep the serve. The world's professional tennis players in front of the world rankings, most have very excellent serve skills. Two different footwork techniques in tennis serve used by most professional tennis players are the foot-up and foot-back serve technique. Most researchers investigated the differences between these footwork techniques using 2-D kinematics data (Elliott et al, 1983). However, little evidence has demonstrated that which serve technique is better (Bylak et al, 1998) or if a difference exists between foot-up and foot-back technique using 3-D analysis (Elliott et al, 1996). The purposes of this study were to investgate the differences in 3-D kinematics between the foot-up and foot-back tennis serve techniques.

METHOD: Eight tennis players (height: 181.5±6 cm, weights:74.9±6 kg, age: 21.5 ±2 years and experience for tennis: 11.1±2.8 years) participated in this study. Vicon Motion System with 10 cameras (frequency 400 Hz) was used to capture the upper extremity (U/E) and lower extremity (L/E) movements of the tennis player during serving with the dominant hand. Fifty-three reflective markers were attached on the anatomic landmarks. In addition, five markers were also adhered on the tennis racket (Figure 1). Visual 3D was used to analyze the collecting data of dominated hand. Ball velocities were also recorded. Three phases (preparation phase, loading phase, and swing phase) were defined. The 'preparation phase' begins when the subject initiates his motion and ends when the racket reaches the highest level. The 'loading phase' begins when the racket reaches the highest level and ends when the racket achieves the most posterior position. 'Swing phase' begins when the racket achieves the most posterior position and ends when the subject completes the serve. Two-sample t-test test was used to identify statistically significant differences in ball velocity and joint angles and angular velocities between two serving techniques. All data analyses were performed using SPSS for window 12.0 (SPSS, Inc., Chicago, IL).

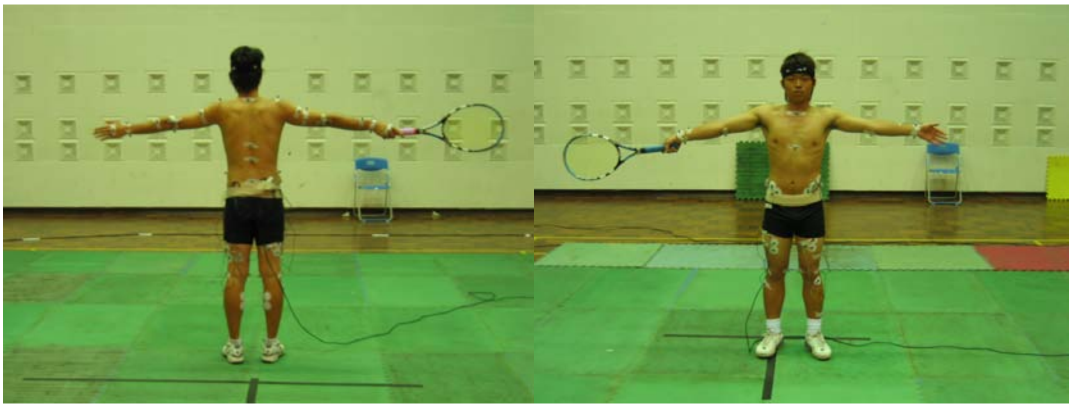

Figure 1. Markers were attached on right and left forehead and posterior head, sternal notch, processus xiphoideus, processus spinosus of the 7th cervical and 8th thoracic vertebra, most dorsal point on the acromioclavicular joint, upper arms, medical and lateral epicondyles, ulnar styloid, radial styloid, 2nd and 5th metacarpal joints, sacrum, anterior superior iliac spine, mid-thigh-cuff with marker on wand, lateral femoral epicondyle, mid-shank-cuff with marker on wand, lateral malleolus, second metatarsophalangeal joint, heel.

RESULTS: The preparation phase was about 65% of serving phase, while the loading phase and the swing phase were 25% and 10%, respectively. Ball velocities during foot-up serve and foot-back serve were listed in Table 1. Ball velocity during foot-up serve was faster than that during foot-back serve. The differences of ball velocities were less than 5 km/hr during different serving. The four types of tennis serving skills were similar in U/E joint angles and angular velocities.

Table1. Ball velocities (km/hr) of each subject including mean and standard deviation (SD) values during four types of tennis serving (RFU: Foot-up serving to the right court; RFB: Foot-back serving to the right court; LFU: Foot-up serving to the left court; LFB: Foot-back serving to the left court)

No. subject	RFU	RFB	LFU	LFB
1	179	183	184	183
2	172	166	174	167
3	184	178	176	162
4	187	191	189	190
5	166	172	174	170
6	178	165	172	173
7	185	178	186	176
8	170	171	168	170
Mean	177.62	175.5	177.87	173.87
SD	7.65	8.79	7.49	8.99

DISCUSSION: Ball velocity during foot-up serve was faster than that during foot-back serve, however, the differences of ball velocities were less than 5 km/hr during different serving. Similar movement patterns (especially U/E) were observed during foot-up and foot-back serve techniques. It appears that the same ball velocities, U/E joint angles and joint velocities were obtained using two different tennis serve techniques.

CONCLUSION: No differences of ball speed, serving phase timing, and kinematics data were found during different tennis serving.

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GROUND REACTION FORCES, KINEMATICS, AND MUSCLE ACTIVATIONS DURING THE SOFTBALL PITCH

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Research has shown that ground reaction forces and lower extremity muscle activation are essential elements to the windmill softball pitch. Therefore, the purpose of this study was to quantify ground reaction forces, motion kinematics and muscle activations during the phases of stride foot plant to ball release of the windmill pitching motion. Ten female windmill softball pitchers participated. Fastballs for strikes were thrown and data revealed that as the windmill softball pitcher had increased ball velocity their vertical ground reaction forces also increased. As the medial forces increased the stride leg gluteus maximus had decreased activation. Proper conditioning of the lumbopelvic-hip complex, including the gluteals, is essential for injury prevention.

KEYWORDS: lower extremity, sEMG, softball pitching, segmental sequencing, vertical ground reaction force

INTRODUCTION: The sport of fast-pitch softball has become very popular for female athletes. It has been reported that during 2003 more than 2 million girls participated in the sport of softball (Guido et al., 2009). Despite its popularity, research regarding the sport has been limited. Previously, ground reaction forces have been examined in softball pitching because of the importance of the lower extremity throughout the pitching motion (Werner et al., 2005). Recently, Guido and colleagues (2009) reported on ground reaction forces and throwing mechanics in youth windmill softball pitchers and found during stride foot contact forces are generated anteriorly, medially, and vertically in attempt to break the forward momentum of the body and provide a base of support for trunk rotation and ball release. Data are yet to be reported on ground reaction forces, lower extremity kinematics, and muscle activations during the windmill softball pitch. Therefore the purpose of the study was to quantify ground reaction forces, motion kinematics, and muscle activations during the windmill softball pitch. It was hypothesized that there would be characteristic ground reaction forces, kinematics, and muscle activation patterns and that variations would be indicative of pitching velocity.

METHODS: Ten female windmill softball pitchers (17.6 ± 3.47 years, 166.9 ± 7.0 cm and 67.4 ± 12.2 kg) volunteered to participate in the study. All participants had recently finished their competitive high school spring softball seasons, and were deemed appropriately conditioned for participation. All testing protocols used in the current study were approved by the University's Review Board.

Participants reported for testing prior to engaging in resistance training or any vigorous activity that day. Location of the right and left gluteus maximus as well as the right and left gluteus medius were identified through palpation. Identified locations for surface electrode placement were shaved, abraded and cleaned using standard medical alcohol swabs. Subsequent to surface preparation, adhesive 3M Red-Dot bipolar surface electrodes (3M, St. Paul, MN) were attached over the muscle bellies and positioned parallel to muscle fibers using techniques described by Basmajian and Deluca (1985). Once all electrodes had been secured, manual muscle tests (MMT) were conducted using techniques described by Kendall et al.(1993). Manual muscle testing was conducted to establish baseline readings for each participant's maximum voluntary isometric contraction (MVIC) to which all sEMG data would be compared.

Surface electromyographic (sEMG) data were transmitted to The MotionMonitor™ motion capture system (Innovative Sports Training Inc, Chicago IL), through a Noraxon Myopac 1400L 8-channel amplifier. The signal was full wave rectified and smoothed based on the smoothing algorithms of root mean squared at windows of 100 ms. Throughout all testing, sEMG data were sampled at a rate 1000 Hz. In addition, all sEMG data were notch filtered at frequencies of 59.5 Hz and 60.5 Hz respectively (Blackburn and Pauda, 2009).
Kinematic and kinetic data were collected using The MotionMonitor™ motion capture system (Innovative Sports Training, Chicago IL). Prior to completing test trials, participants had a series of ten electromagnetic sensors attached at the following locations: (1) the medial aspect of the torso at C7; (2) medial aspect of the pelvis at S1; (3) the distal/posterior aspect of the throwing humerus; (4) the distal/posterior aspect of the throwing forearm; (5) the distal/posterior aspect of the non-throwing humerus; (6) the distal/posterior aspect of the non-throwing forearm; (7) distal/posterior aspect of stride lower leg; (8) distal/posterior aspect of the upper stride leg; (9) distal/posterior aspect of non stride lower leg; and (10) distal/posterior aspect of non stride upper leg (Myers et al., 2005). Following the attachment of the electromagnetic sensors, an eleventh sensor was used to digitize the bony landmarks.
Following all set-up and pre-testing protocols, participants were allotted an unlimited time to perform their own specified pre-competition warm-up routine. During this time, participants were asked to spend a small portion of their warm-up throwing from the indoor pitching surface to be used during the test trials. After completing their warm-up and gaining familiarity with the pitching surface, each participant threw a series of maximal effort fastballs for strikes toward a catcher located the regulation distance (12.2 m). The pitching surface was positioned so that the participant's stride foot would land on top of a 40 x 60 cm Bertec force plate (Bertec Corp, Columbus, Ohio) which was anchored into the floor. For the current study, those data from the fastest pitch passing through the strike-zone were selected for detailed analysis. Pitch velocity was determined by JUGS radar gun (OpticsPlanet, Inc., Northbrook, IL) positioned at the base of the pitching surface and directed towards home plate.
Raw data regarding sensor orientation and position were transformed to locally based coordinate systems for each of the respective body segments. Euler angle decomposition sequences were used to describe both the position and orientation of the torso relative to the global coordinate system (Wu et al., 2002; Wu et al., 2005). The use of these rotational sequences allowed the data to be described in a manner that most closely represented the clinical definitions for the movements (Myers et al., 2005).
Data were analyzed in the current study using the statistical analysis package SPSS 15.0 for Windows. Data for the fastest strike mean and standard deviation, for all sEMG and kinematic and kinetic parameters were calculated. Phase 1, was defined as the start of pitching motion to the top of back swing (TOB). Phase 2, was from TOB to stride foot plant (SFP). Phase 3 was from SFP to ball release (BR), and Phase 4 was from BR to the completion of follow through.

RESULTS: Means and standard deviations of kinematic, kinetic and sEMG data are presented in Figures 1-3. Average ball velocity reported was 24.1±1.38 m/s.

DISCUSSION: This is the first study to investigate ground reaction forces, kinematics and sEMG of the windmill softball pitch. It is evident that the participants in the current study generated large breaking forces and vertical forces to drive toward the plate in order to generate the greatest ball velocity (Figure 1). Breaking forces revealed on the average 35.9%±10.3%BW for all participants. However, when examining stride length, those with greater stride length, as defined by TOB occurring prior to SFP, demonstrated breaking forces of 31.5%±12.5%BW compared to 41.4%±8.5%BW respectively. When examining vertical forces, the average was 179%±38.2%BW; however, those with longer strides did exhibit greater vertical forces than those with shorter stride lengths. It has been reported that those with longer stride lengths have greater ball velocity (Guido at al., 2009); however, in the current study those with longer stride lengths had on average 0.51m/s slower velocity than those with shorter stride lengths. Those with higher velocities exhibited higher vertical ground reaction forces than those with lower velocities.

Gluteal muscle activations of the stride leg during Phase 3 of the softball pitching motion demonstrated pelvis stabilization and torque generation in preparation for ball release (Figure 2). Typically the gluteal muscle group provides stabilization when on single leg support. Pelvic stabilization is important for efficient energy transfer up the kinetic chain from the hips to the pelvis and scapula on to the shoulder, elbow, hand and wrist for ball release. This action is evident by the presented relationship of the non stride leg gluteus maximus having greater activation in those with greater ball velocities. In addition, as the medial (Fz) forces increased, stride leg gluteus maximus had decreased activation. This is important to note in that as there were greater medial forces at the lower extremity on the stride leg, the stride leg gluteus medius had decreased activation. The gluteus maximus acts to extend the hip and then allows for external rotation of the hip. The decreased activation of the gluteus maximus on the stride leg indicates that it was not as active in externally rotating the stride hip thus resulting in increased medial ground reaction forces. The lack of gluteal activity was evident by the great amount of stride knee abduction at foot contact (Figure 3).

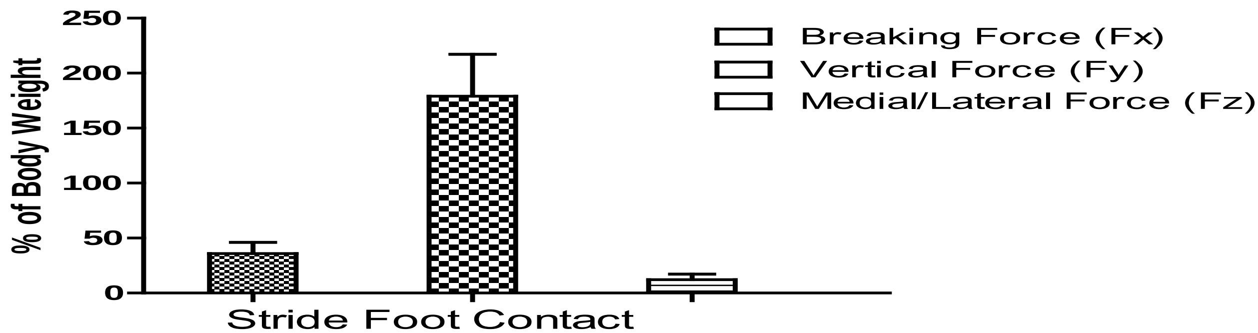

Figure 1. Means and standard deviations of ground reaction force magnitude at stride foot contact presented as a percent of the participant's body weight.

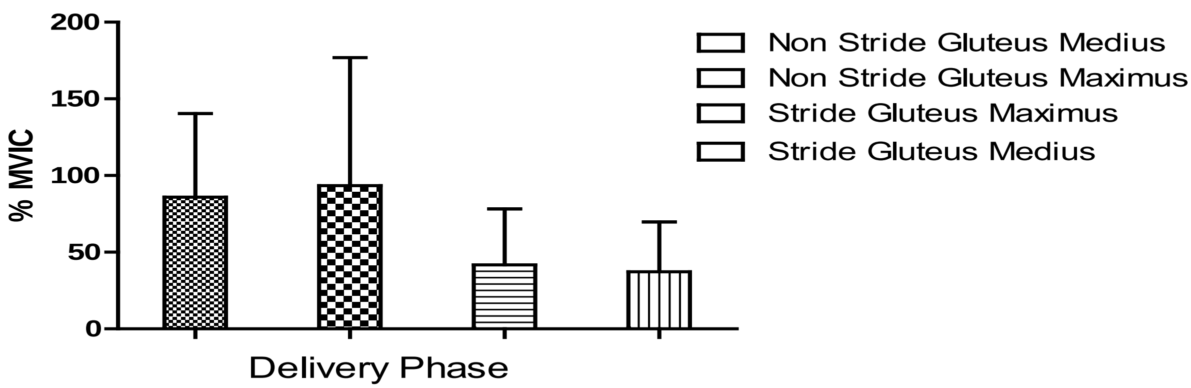

Figure 2. Means and standard deviations of the gluteal musculature, as presented as a percent of their MVIC, during the delivery phase of the windmill pitching motion.

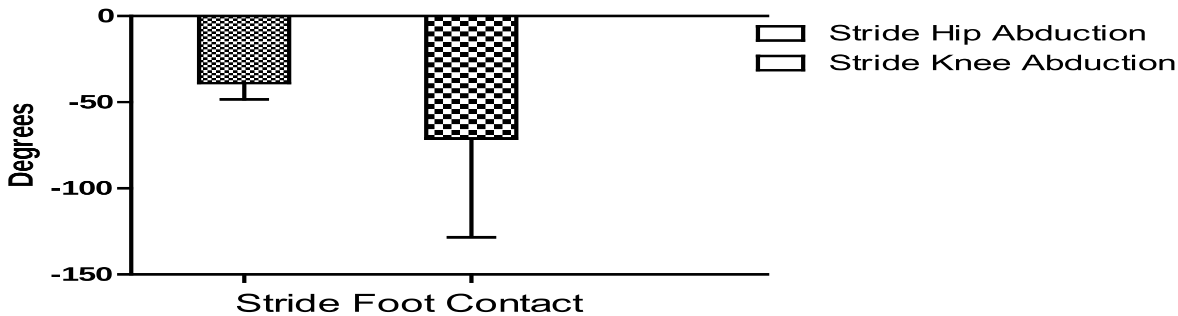

Figure 3. Means and standard deviations of lower extremity kinematic parameters of the stride leg hip and knee as presented in degrees experienced at stride foot contact.

CONCLUSIONS: Data revealed that as the windmill softball pitcher increased ball velocity their vertical ground reaction forces also increased. Thus, further studies are warranted examining the muscle activations of the lower extremity and their importance in the effectiveness of the windmill softball pitch. However, from the data presented, it is evident that strength and conditioning of the gluteal muscle group bilaterally is salient in the windmill softball pitch.

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ELECTROMYOGRAPHIC FACTORS CORRELATED WITH SOFTBALL BATTING PERFORMANCE

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KEYWORDS: ball exit velocity, bat head velocity.

INTRODUCTION: Specific muscle strength and coordination are required to produce a successful hit after a fast downswing in response to a high-velocity ball (Katsumata, 2007). Electromyographic (EMG) recording during the muscle contraction was a measurement to realize the motor strategy of sports. Several EMG researches about baseball or softball pitching have been documented in literature (Saito et al., 2001). However, there was very little research about softball batting surveyed in literature. Therefore, the purposes of this study were to investigate the muscle activities in upper extremities during softball batting and to analyze if any muscle activity is correlated with the ball exit velocity and bat head velocity.

METHOD: Seventeen female college softball batters participated in this study. Surface EMG signals on anterior deltoid, posterior deltoid, pectoralis major, biceps brachii and triceps brachii were measured bilaterally during softball batting. In this study, pushing arm was defined as the upper limb in the same side of the supporting leg while leading arm was defined as the upper limb in the same side of the leading leg. MA300 EMG system (Motion Lab Systems, Inc.) was used at a sampling rate of 1000 Hz. Maximum voluntary contraction (MVC) was measured for each muscle before collecting batting trials.
VICON 612 motion analysis system (Oxoford Metrics Limited.) with six digital cameras was used. The sampling rate was 250 Hz. In softball batting trials, the trajectories of the softball wrapping with reflective material and the reflective marker on bat head were measured and EMG signals in upper limbs were simultaneously collected. A softball was placed on a stationary framework. The height of the framework could be adjusted to match the most proper batting position for each subject, about the level of anterior superior iliac spine. Three bats were used, consisted of light (22 oz), regular (26 oz) and heavy (30 oz) bats. Six successful batting trials in which there was proper bat contact were selected for each bat. The maximum ball exit velocity (the batted ball velocity) and the bat head velocity were calculated. The EMG signals in acceleration phase of softball batting were computed as root mean square values and normalized by MVC for each muscle. The average of six trials was calculated for each subject. Pearson correlation coefficient was analyzed to see if any EMG parameter was correlated with softball performance (n=17).

RESULTS: The ball exit velocity and bat head velocity were shown in Table 1. The correlation coefficients with significance ($p<0.05$) were shown in Table 2. Ball exit velocity was correlated with bat head velocity. The heavier bat had the trend of stronger correlation between ball velocity and bat velocity.
With the use of a light bat (22 oz), posterior deltoid in leading limb (r=0.588) and triceps brachii in pushing limb (r=0.495) were correlated with bat head velocity. With the use of a regular bat (26 oz), posterior deltoid in leading limb was correlated with ball exit velocity (r=0.551), posterior deltoid in pushing limb (r=0.520) and biceps brachii in leading limb (r=0.484). Triceps brachii in pushing limb was correlated with anterior deltoid (r=0.643) and posterior deltoid in pushing limb (r=0.622). With the use of a heavy bat (30 oz), posterior deltoid in pushing limb was correlated with biceps brachii in leading limb (r=0.811), anterior deltoid in pushing limb (r=0.559) and posterior deltoid in leading limb (r=0.513).

Table 1. The ball exit velocity and bat head velocity (m/s)

Bat weight	Ball exit velocity	Bat head velocity
22 oz	20.54±1.81	25.02±2.21
26 oz	24.72±1.48	24.62±2.27
30 oz	20.64±2.58	24.23±2.50

Table 2. Correlation Coefficients between ball exit velocity, bat head velocity and EMG parameters (L: leading arm; P: pushing arm)

Bat weight	Variable 1	Variable 2	r	p
22 oz	Ball exit velocity	Bat head velocity	0.652	0.005
	Bat head velocity	Posterior deltoid (L)	0.588	0.013
	Bat head velocity	Triceps brachii (P)	0.495	0.043
26 oz	Ball exit velocity	Bat head velocity	0.804	0.001
	Anterior deltoid (P)	Tricpes brachii (P)	0.643	0.005
	Posterior deltoid (P)	Triceps brachii (P)	0.622	0.008
	Ball exit velocity	Posterior deltoid (L)	0.551	0.022
	Posterior deltoid (L)	Posterior deltoid (P)	0.520	0.032
	Posterior deltoid (P)	Anterior deltoid (P)	0.515	0.034
	Posterior deltoid (L)	Biceps brachii (L)	0.484	0.049
30 oz	Ball exit velocity	Bat head velocity	0.832	0.000
	Posterior deltoid (P)	Biceps brachii (L)	0.811	0.000
	Posterior deltoid (P)	Anterior deltoid (P)	0.559	0.020
	Posterior deltoid (P)	Posterior deltoid (L)	0.513	0.035
	Posterior deltoid (L)	Anterior deltoid (L)	-0.574	0.016

DISCUSSION: Bat head velocity was highly correlated with ball exit velocity (all bats), implying that a fast bat head velocity is useful for performing a high ball exit speed in softball batting. Also, the posterior deltoid in leading arm and triceps brachii in pushing arm were correlated with bat velocity or ball velocity. The importance of these two muscles in softball batting might be obviously revealed. The batting skills might be improved if the motor strategy of upper extremity could be modified as highlighting the activations of posterior deltoid in leading arm and triceps brachii in pushing arm.

CONCLUSION: The findings of this study demonstrated the specific EMG in the muscles of leading arm and pushing arm, which showed significant correlation with bat velocity and ball velocity. The correlation of muscle activation with bat velocity suggests that specific muscle conditioning programmes may improve batting performance.

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INSIGHTS OF TAKE-OFF OF GROUND REACTIONS FORCE IN HIGH JUMP

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KEYWORDS: High Jump, Take-off, Ground Reaction Forces.

INTRODUCTION: The take-off plays an important role on the high jump (HJ) performance, since it is there where the trajectory of the centre of mass is defined (Tellez, 1993) in order to achieve the maximum possible height and clear the bar (Dapena, 1988). The literature has presented several studies using kinematic parameters, however there are very few studies using dynamic parameters to evaluate the take-off (Coh & Supej, 2008).
In contrast to other jumps, the HJ presents an increased complexity due to the curvilinear approach-run pathway, imposing a three-dimensional analysis of the movement. On the other hand, the point where the jumpers perform the take-off differs considerably for different subjects. For those reasons research presented in the literature is fundamentally based in the kinematics of the approach-run and take-off. Although great knowledge was achieved in the kinematic of HJ, very few studies can be found concerning the ground reaction forces (GRF). However the knowledge of the GRF pattern enables a more thorough evaluation of the jumper technique and efficiency. The purposes of this study were (i) to evaluate the HJ take-off technique using force plate data based on the characterization of the GRF pattern and (ii) definition of take-off GRF profiles as function of technical and conditional characteristics of different athletes.

METHOD: Nine male high jumpers from different levels (age: 18.7±3.7 years old; height 1.81±0.07 m; body mass: 71.3±6.7 kg, personal record: 1.91±0.2 m) took part on this study. A strain gauge force plate (Bertec 4060-15) placed on the take-off point and sampling at 1000 Hz was used to collect the GRF data during HJ take-off. The force plate was synchronized with a high-speed camera sampling at 1000 Hz by photocells placed 5 m before the take-off point. Each jumper performed the trials in accordance with the official rules of HJ competition. GRF data was treated in software developed in Matlab environment. Firstly the GRF was corrected by a rotation matrix taking into consideration the foot angle relatively to the HJ bar. Then, global and local maximum and minimum in each component of GRF, their time of occurrence, impulses and contact time were determined. Finally data were resampled over new time vector between [0; 1000] and scaled to their maximum value to obtain GRF profiles for each jumper during take-off.

RESULTS: The maximum heights that the jumpers cleared were between 1.75 and 2.05 m with a mean value of 1.83±0.13 m. The mean take-off time was 0.197 s±0.02 s, with the minimum and maximum values between 0.169 s and 0.228 s, respectively. A mean angle of 40.5±10.2º was observed between the foot and the HJ bar. The vertical GRF varied between 1170 N and 4110 N. We found two different vertical GRF profiles, one with two peaks, and other with three peaks. The peak values in this component for athletes with two peaks was 2920.0±371.2 N, and 3027.8±445.9 N, while for those with three peaks, 3400.8±404.7 N, 3118.1±460.7 N and 2605.2±257.6 N. Antero-posterior forces ranged from -2418 N for the breaking phase and 152 N for the propulsive phase. The profile of this component presents only negative values. The medio-lateral forces varied between 1026 N and -589 N.

DISCUSSION: Results obtained for GRF are much lower than those presented in the literature, probably because our sample is composed by young level athletes with only two national elite level jumpers.
In contrast to other studies, the GRF were corrected and changes were perceived in the magnitude of the antero-posterior component, and in the pattern of the medio-lateral component, while the vertical component was kept unchanged.
Our results show that some features resulting from the approach-run can be observed in the take-off. When the athletes perform a correct curve the forces in the antero-posterior component were always negative towards inside the curve which means that the jumper do not generate additional force to go outward (into the mat). Motions from the leading leg and arms were evaluated in the medio-lateral component, identified by the change in the sign of the force. Although the magnitude intensity of the forces is low, they are important to understand how the jumper is managing its free limbs. The actions of the take-off leg can be observed particularly in the vertical GRF. We noted that when the time of occurrence of maximum and minimum points particularly in the vertical and medio-lateral components matched there is a strong association with an effective take-off action.
Concerning the GRF profiles we found two different kinds of profiles. Figure 1 represents the profile with two peaks in the vertical GRF, while in Figure 2 it is shown the three peaks profile. The most reactive athletes show two peaks in the vertical component and, contrary to the other profile, show a deep increase of forces in the last peak. The antero-posterior component is similar in all the profiles. With regard to the medio-lateral component, large variation is found. These results allowed the determination of subject specific take-off profile. Each one reflects the technical and conditional characteristics of a given athlete. If the sample was larger it is likely that more different profiles would be found. In the future, GRF should be combined with kinematics in order to obtain more insights about the behaviour of GRF in HJ.

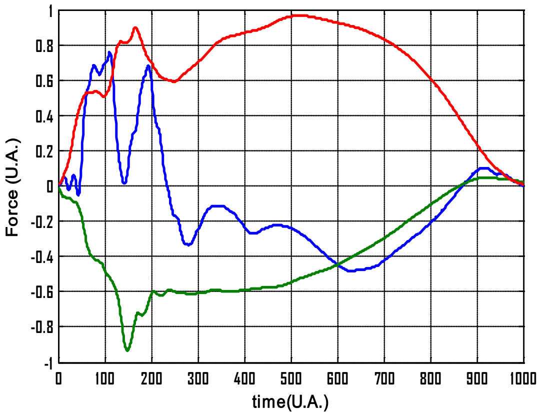

Figure 1. Take-off profile of ground reactions forces of athlete S1. Mean curves from different trials.

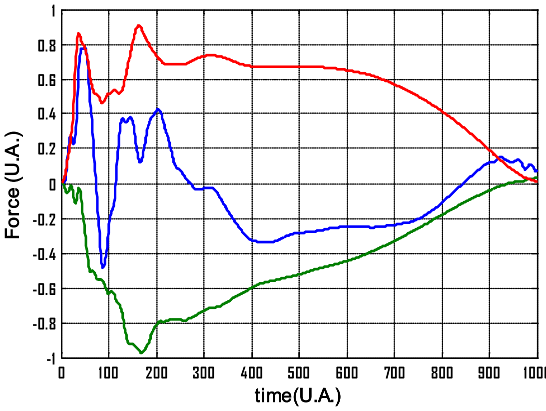

Figure 2. Take-off profile of ground reactions forces of athlete S6. Mean curves from different trials.

CONCLUSION: The main conclusions of this work are that GRF can be used to perform technical evaluation and that each jumper has specific take-off characteristics resulting from its technical and conditional characteristics.

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KINEMATICAL PARAMETERS CONTRIBUTION TO THE FLIGHT HEIGHT USING ONE-FOOT OR TWO-FOOT TAKE-OFF

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The purpose of this study was to investigate which of the kinematics parameters most contribute to vertical flight heights. Eight subjects were filmed using Sony digital camera with 25 images from the sagittal plane during execution of the following vertical jump conditions: free two-foot take-off, free one-foot take-off, fixed arm two-foot take-off, and fixed arm one-foot take-off. Arm swing contribution, leg swing contribution, height of centre of gravity at take-off (HCGTO), vertical velocity at take-off, arm's angular momentum, work, and power were analyzed in each condition using stick figures according to Clauser, McConville, and Young (1969). Correlation and regression analysis indicated that HCGTO contributed the most to the flight heights in all conditions.

KEY WORDS: one-two-foot, vertical jump, kinematics.

INTRODUCTION: In vertical jumps, two components contribute to the power output: the amount of optimum impulse generated at take off and type of muscle contraction with related mass. So it is assumed that having more leg muscle mass would be associated with a better utilization of elastic component in the muscle and achievement of a higher jump height (Kilani et al., 1989). Thus, a two-foot take off should produce a higher vertical jump than a one-foot take off since the jumper carries the same body weight. Vint and Hinrichs (1996) quantified the differences between one- and two-foot vertical jumping performances and they reported greater flight heights during two-foot jumps, as expected, but the magnitude of this difference was only about 9 cm. One-foot jumps might benefit from an increased take off height which is largely attributable to the elevation of the free swinging leg. Factors associated with the development of muscular tension, such as vertical velocity at touchdown during eccentric contraction, height of center of gravity at take-off and horizontal approach velocity may account for the small differences in flight height between one-foot and two-foot jumping performances. The purpose of this current research was to test differences in heights of a vertical jumping performance using one-foot or two-foot take-off and to investigate which of the kinematics parameters are most contributing percent wise to vertical flight heights.

METHOD: Eight subjects with a mean height of 176.62 cm and a mean weight of 67.62 kg were intentionally selected from the physical education students at Sultan Qaboos University in Muscat, Oman (table 1). They were filmed using Sony digital camera (DCR-SX41/R/Max Shutter Speed 1/4000 sec) with 25 images from the sagittal plane during the executions of the following vertical jump conditions free two-foot take-off (FTFT), free one-foot take-off (FOFT), fixed arm two-foot take-off (FATFT), and fixed arm one-foot take-off (FAOFT).

Table 1 Subjects heights and weights

Parameters	Means	SD
Height cm	176.62	8.10
Weight kg	67.62	8.35

Before data collection, each subject signed a consent form and visited the lab once, several days before the measurement and was instructed how to perform each of the jumps. The starting position for the FTFT and FOFT jumps was standing upright with the arms down at the side. For the FATFT and FAOFT jumps, the subjects stood upright with their hands on their hips. The subjects performed each jump maximally 3 times and were given enough rest so that they didn't feel any fatigue from the previous jump conditions. No instructions were given with regard to the amount of knee bend a subject should have. Variables were analyzed manually from the flat Sony TV screen for the following independent variables in each condition using stick figures as noted by Clauser, McConville, and Young (1969): 1) Arm swing contribution (ASC) which is measured as a percent from the maximum height achieved without constraint relative to each condition; 2) leg swing contribution (LSC) which is measured as a percent from the maximum height achieved without constraint relative to each condition; 3) height of centre of gravity at take-off which is the distance measured from the instance of take off to the ground (HCGTO); 4) vertical velocity at take-off (VVTO) which is measured as the displacement between 2 images over time at the instance of take off; 5) arm's angular momentum (AAM) which is measured as the arm mass times the angular velocity of the shoulder flexion to the take off; 6) knee extension phase (KEP) which is the distance measured from the moment of maximum knee flexion to maximum knee extension prior the instance of take off; 7) work (W) which is calculated as body weight times the vertical distance measured from knee extension until the maximum height of CG; and 8) power (P) which is measured by the work out put over time. The dependent variable was the maximum flight height jump achieved in each condition and was measured as the distance between highest points reached of the CG to the ground. Correlations, comparisons, and regression analyses were conducted using SPSS software package version 15.

RESULTS: The means and standard deviations for the independent and dependent variables are presented in Table 2. The power and work output for FTFT and FOFT are also in Table 2. Arms and legs contributions to FTFT and FOFT jumps are displayed in Table 3.

Table 2. Means & standard deviations for FTFT and FOFT

Parameters	Maximum flight height(m)	HCGTO (m)	KEP (m)	VVTO (m/s)	AAM (Kg.Rad/S)	Work (kg.m)	Power (kg.m/s)
Mean	1.61	1.06	0.39	2.06	7.75	127.24	1485.1
Standard deviation	0.10	0.05	0.06	0.94	2.36	30.58	346.8

Table 3. The contribution in percentages for the arms and legs in jump conditions.

Jump conditions	Free jumps		Fixed Arm Jumps		Difference mean	Arm's Contribution %	Leg's Contribution %
	Mean	SD	Mean	SD			
Two-foot	1.16	0.68	1.11	0.068	0.05	4.31%	95.69%
One-foot	1.14	0.50	1.12	0.057	0.02	1.75%	98.25%

Table 4. The percentages of the parameters contributed the most to the flight height in the 3 conditions FOFT, FTFT, and FAOFT as it was derived from the linear regression equations

Jump positions	FOFT	FTFT	FAOFT	
Parameters	KEP (m)	HCG before takeoff	Vertical Distance during preparatory phase (M)	Work
Standard deviation	0.04	0.05	0.042	23.26
Constant	1.701	0.294	1.50	
Correlation Coefficient	2.132	0.284	2.522	0.0020
(F) value	**29.82	*50.245	**49.18	
Significance Level	0.002	0.000	0.001	
% Partial Contribution	٪ 83.2	٪89.3	٪75.6	٪75.6
% Cumulative Contribution	٪ 83.2	٪89.3	٪19.5	٪95.1

* F value in (P ≤0.05) = 5.79

** F value in (P ≤0.01) = 13.27

Table 4 illustrates the percentages parameters that contributed the most to the flight height in the 3 conditions FOFT, RTFT and FAOFT.

DISCUSSION: Results derived from the statistical manipulation showed that arms contributed greater flight heights in FTFT than FOFT. This result agrees with that of Holvoet, Lacouture, and Duboy (1999) where arm's swing contributed about 20% at vertical jump.The swinging free leg contributed in one-foot take-off for the height by increasing the CG elevation at take off. Consequently, it increased the momentum of the body in the vertical direction. Researchers referred this to the extreme dynamic range of swinging free leg which was achieved in FOFT and this case explains the results found by Caroline & Wooden I (2000). HCGTO contributed the most to the flight heights in all conditions. (F) Value for both parameters HCGTO & VVTO was significant at the level of (P ≤ 0.01). As it is shown in table 3, the small differences in the flight height between one-foot and two-foot jumping performances may be due to the swinging leg that affect CG elevation at take off with increased vertical velocity which counteracts the two-foot muscle force production against the same body weight carried (Vint & Hinrichs, 1996).

CONCLUSION: The powerful swings of arms, the high strength of hip joint with short downward motion (KEP) of CG are much more helpful to improve the flight jump height. It is recommended that training programs take into account the kinematics that contributes to the vertical jumps. In addition, it is helpful to practice with both feet and single foot jumping drills without the aid of swinging neither legs nor arms if muscular strength is needed. If power on the other hand is needed, swinging arms and free leg are crucial to maximize performances.

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RELIABILITY OF ACUTE STATIC STRETCH IMPACT ON VERTICAL JUMP HEIGHT

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The purpose of this study was to examine the reliability of the acute effect of stretch on vertical jump performance. Twenty-four subjects completed eight trials. In each trial warm-up, three pre-stretch jumps, stretching, and three post-stretch jumps were completed. Intraclass correlation coefficients (ICC) were used to evaluate the reliability of the stretch impact on vertical jumping across trials. While pre-stretch and post-stretch jumps were highly reliable (ICC=0.99), the difference from pre-stretch to post-stretch was not (ICC=0.07). The stretch impact on vertical jump may not be reliable for subjects across trials.

KEY WORDS: intraclass correlation coefficient, consistency

INTRODUCTION: Many investigations examining the acute impact of static or proprioceptive neuromuscular facilitation (PNF) stretching on vertical jumping have been conducted. Many have found acute stretching to decrease vertical jump performance (Behm et al., 2006; Bradley et al., 2007; Church et al., 2001; Hough et al., 2009; Walter & Bird, 2009), while others found stretching to have no significant impact on subsequent vertical jump performance (Knudson et al., 2001; Unick et al., 2005; Young & Elliott, 2001). Even with such varied findings, many have concluded that stretching immediately prior to performance has little benefit and potential significant reductions in performance and should be avoided.

While movement outcomes often have declined after stretching, investigations of other related biomechanical variables have not produced consistent results. Previously some researchers have attributed changes in performance after stretch to musculotendon stiffness (Guissard et al., 2001; Toft et al., 1989). Yet others have not supported the idea of musculotendon stiffness (Goodwin et al., 2009; Knudson et al., 2001; Nelson et al., 1996). Investigations of related kinematics (Knudson et al., 2001), flexibility (Behm et al., 2006), muscle activity (Hough et al., 2009), and ground reaction forces (Walter & Bird, 2009) failed to find variables associated with reduced vertical jump height immediately after stretching.

The acute impact of static and PNF stretching on vertical jump height and its associated variables has eluded researchers for some time. Investigators have reported a strong reliability (r = 0.86 to 0.99) of individual measures such as jump height, power, and ROM (Behm et al., 2006; Bradley et al., 2007; Walter & Bird, 2009), yet no one has reported the reliability of the stretch impact on vertical jump performance. Knudson et al. (2001) did not find a significant impact on vertical jump performance but did report the stretch impact lacked uniformity across subjects when 55% of the participants decreased in jump height, 10% had no change, and 35% of subjects increased after stretch. Walter and Bird (2009) found a significant impact of stretch on vertical jump, but unlike Knudson et al. (2001), 73% of subjects decreased in jump height, 15% had no change, and 12% increased in jump height. While the characteristics of the subjects in each study may have influenced who was impacted and who was not, there is no evidence to support the reliability of the acute impact of stretching on vertical jump height. The purpose of this study was to examine the reliability of the acute effect of static stretching on vertical jump performance.

METHODS: Prior to participation 13 female and 11 male recreational and varsity athletes (age: 19.6±0.9 yrs, height: 1.75±0.12 m, and weight: 76.0±15.6 kg) from the university population provided informed consent in accordance with testing procedures approved by the university Institutional Review Board. Each of the 24 subjects completed eight sessions, separated by a minimum of 24 hours. Each data session required the subject to warm-up on a Monark cycle ergometer at 120 W for five minutes, execute 2-4 warm-up vertical jumps, then complete three jumps, stretch, and immediately complete three more jumps. Jumps were performed with both hands on hips to isolate the lower extremities and to eliminate potential confounding arm coordination; subjects were instructed to jump as high as possible for all jumps. A Kistler force plate (model 9286AA) was used to measure the ground reaction force data (1000 Hz sampling rate).

After the initial set of three jumps, subjects completed four static stretches for the gluteal, hamstring, quadriceps, and calf muscle groups. Each stretch was performed once unilaterally and held for a timed thirty seconds, as described in the National Strength and Conditioning Association guidelines (Baechle & Earle, 2000). Each stretch was completed on both sides of the body. Adequate stretch was defined as committing to full range of motion at each joint until slight discomfort, but not pain, was achieved. Stretching techniques were demonstrated to the subjects prior to data collection to ensure subject understanding of proper technique, and monitored during the experimental stretching routine.

For the unilateral gluteus stretch subjects sat with knees flexed and their feet flat against the floor. After crossing one leg over the thigh of the other leg, they grasped the back of that same thigh with both hands. Subjects pulled their legs towards their torso to stretch. For the unilateral seated hamstring stretch subjects sat with an anterior tilt of the pelvis. The leg being stretched remained outstretched while the uninvolved leg was flexed in a figure-four position. Subjects then were instructed to lean forward, flexing at the hip, and reach with their hands towards their toes. For the unilateral standing quadriceps stretch subjects stood on one leg with a posterior pelvic tilt and one hand against a wall for balance. Subjects grasped non-weight bearing foot, bringing the knee into flexion as far as possible while keeping the knee perpendicular to the floor. For the unilateral standing calf stretch subjects stood with both hands placed against a wall in front of them. While keeping left knee slightly flexed, subjects were to move right foot back about half a meter and place right heel and foot flat on the floor.

Vertical jump height was calculated from the time off the force plate. Reliability for the pre-stretch jumps, post-stretch jumps, and stretch impact were evaluated using Intraclass Correlation Coefficients (ICC) across the eight trials. Based on jump reliability for consecutive jumps (ICC = 0.97), only the second pre-stretch and post-stretch jump heights were used to determine the effect of stretching on vertical jump. Sit and reach measures completed after the first and eighth sessions were compared with a paired-samples t-test. A repeated measures ANOVA was used to compare stretch impacts across trials and to evaluate gender differences.

RESULTS: The changes in jump height did not significantly ($p>0.05$) affect women differently than men. Subsequent analyses collapsed data across gender. Sit and reach significantly ($p<0.05$) increased over the time period of data collection (from 13.8±6.0 cm to 14.5±5.9 cm, $p<0.05$). The effect size was 0.175, reflecting a small change over time. The ICC for all pre-stretch jumps was 0.99 and the ICC for all post-stretch jumps was also 0.99. The ICC for all differences from pre-stretch to post-stretch, an indication of the impact of the stretch, was 0.07. No significant differences were found when comparing the stretch impact on vertical jump heights across trials ($p>0.05$). See Figure 1.

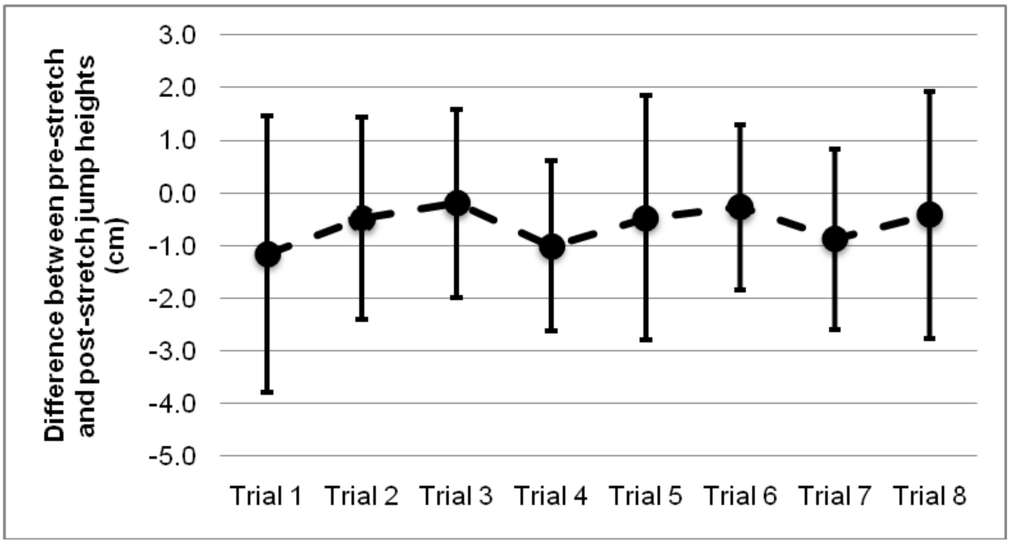

Figure 1: Mean (±SD) differences between pre-stretch and post-stretch jumps for all trials. No differences were found between means. The ICC was 0.07.

DISCUSSION: The purpose of this investigation was to evaluate the reliability of the stretch impact on vertical jump performance. Changes in flexibility over the eight trials were not unexpected but, with an effect size of 0.175, the change was small and potentially inconsequential. Behm et al. (2006) examined flexibility and changes in flexibility on the stretch impact in vertical jump height, but did not find any relationship between flexibility and changes in flexibility and stretch influences of vertical jump performance. The changes were, however, a good reflection of the subject's adherence and effort in the stretching routine.

The pre-stretch and post-stretch ICC values were strong; subjects were highly consistent performers from trial to trial, both before and after stretching. These values were consistent with jump reliability values found in previous research (Behm et al., 2006; Bradley et al., 2007; Walter & Bird, 2009). In contrast to the pre-stretch and post-stretch reliability, the ICC for the changes in jump height attributed to stretch reflected poor reliability. Stretching did not consistently affect subjects across trials. So little consistency of stretch impact meant subjects could not be grouped according to who was "responsive" to the stretch and who was "unresponsive" to the stretch. Further, the trial-to-trial variation may mean any possible negative influence on vertical jump performance would also vary from day to day.

No subject decreased in jump height for all eight trials. For one or more trials each subject improved in jump height after stretching. Though the average change decreased by 0.61±2.01 cm across all eight trials, the impact of stretching on performance varied, but not consistently in a negative or positive manner. The average relative reductions for each trial were between 0.9% and 3.7%; no trial had an average increase in jump height. On average, some trials were more affected by stretch than others. There was a tendency to decrease jump height after stretching, but not always, and not for the same people each time.

The influence of stretch on vertical jump performance did not differ from trial to trial. However, if only the first trial was considered, the stretch would have been evaluated as having a significant ($p<0.05$) effect on jump height, the mean decreased by 3.7%, typical of other studies of stretch impact on jump height (Behm et al., 2006; Bradley et al., 2007; Hough et al., 2009; Walter & Bird, 2009). The inconsistency of stretch impact on vertical jump across trials in these results may explain the lack of consistent impact of stretching in other studies, where significant decreases in jump height were sometimes found (Behm et al., 2006; Bradley et al., 2007; Church et al., 2001; Hough et al., 2009; Walter & Bird, 2009), but not all of the time (Knudson et al., 2001; Unick et al., 2005; Young & Elliott, 2001). When the stretch influence was significant in previous research, it was often attributed to various biomechanical and physiological phenomena (Goodwin et al., 2009; Guissard et al., 2001; Knudson et al., 2001; Nelson et al., 1996; Toft et al., 1989). It seems unlikely that a biomechanical or physiological change would be unreliable; perhaps some other factor is changing the performance of those who stretch.

CONCLUSIONS: If the impact of stretching is not reliable, it is also not valid. If the impact of pre-activity stretching is not reliable, more research is needed to understand what is influencing the athletes' performance after stretching.

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EFFECT OF VERBAL AND VISUAL FEEDBACK ON PEAK TORQUE DURING A KNEE JOINT ISOKINETIC TEST

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The purpose of this study was to assess the effect of verbal and visual feedback on peak torque in male subjects. Thirty male subjects were tested on four separate occasions by executing a knee flexion/extension isokinetic set of four maximal repetitions, at velocities of 60, 120, 180, 240 and 300 deg/sec with a 60 second rest between each velocity set. The velocity order was randomized and visual and verbal feedback to subjects was randomly assigned. A 2 X 5 repeated measures ANOVA was used to analyze the data with $\alpha < 0.05$. There were no significant differences in peak torque regardless of the presence or absence of feedback. The conclusion of this study was that feedback does not increase peak torque during concentric isokinetic testing.

KEYWORDS: isokinetic, peak torque

INTRODUCTION: During isokinetic testing there is also a disparity in whether or not to provide feedback. There have been studies that do not report whether there was feedback (Cramer et al., 2000; Kovaleski et al., 1992; Kovealeski & Heitman, 1993; Parcell et al., 2002; Wilhite et al., 1992), some report verbal encouragement (Tourney-Chollet et al., 2000; Gioftsidou et al., 2006; Ozcaldiran, 2008) and one that reported both verbal and visual feedback (Dauty et al., 2007). This leads to the question of whether feedback is important in isokinetic testing and whether it seems to affect outcome measures, such as peak torque. The importance of knowing the effect of feedback would be important to clinicians who are using this form of evaluation to keep or release patients in rehabilitation. Moreover, other users of isokinetic devices such as athletic trainers or strength and conditioning coaches could be influenced to continue or reduce training protocols based on the peak torque the athletes are exhibiting. In both examples the client may not be working as hard as they could due to lack of motivation which might be exposed by the use of feedback.

Furthermore, much of isokinetic research is conducted using an ascending order of velocity sets (Gioftsidou et al., 2006; Ozcaldiran, 20008; Parcell et al., 2002). However others have used random order of velocities to assess peak torque (Cramer et al., 2000; Dauty et al, 2007; Greig, 2008; Timm & Fyke, 1993; Tourny-Chollet et al., 2000; Wilhite et al., 1992). Kovelski et al., (1992) and Kovaleski and Heitman (1993) investigated changing the order of velocities during training and found that it might be advisable in certain situations to train at a faster speed prior to progressing to slower speeds. Regardless of this research, it seems that much of the published literature regarding isokinetic testing protocols reports the use of ascending order without there necessarily being a scientific foundation to do so.

The purpose of this study was to assess the effect of verbal and visual feedback on peak torque. Secondly, to investigate the effect of velocity order (ascending or descending) on peak torque.

METHOD: Thirty apparently healthy, college-aged male subjects were recruited for this study. Twenty-one participants had been members of a Division III collegiate athletic team during the last two academic years [CA]. Nine participants had not been members of a Division III athletic team during the last two years) [RA]. Subjects were excluded if they had a previous knee injury. The study was approved by the University IRB and all subjects signed informed consent. The mean age, height, and mass of the CA group were 20.76 ± 1.14 yrs, 181.31 ± 9.92 cm, and 85.43 ± 13.01 kg., and for the RA group 20.67 ± . 71 yrs, 180.90 ± 10.06 cm, and 90.76 ± 17.54 kg.

A Cybex NORM isokinetic machine was used for all testing. For the present study gravity correction was integrated in all tests and the Cybex NORM was calibrated prior to collection of any data.

Subjects reported to the lab on six separate occasions. The first two were familiarization sessions and four were experimental testing sessions, all of which included a required five minute warm up on a bicycle ergometer at a self selected pace. The warmup on the isokinetic machine included four concentric submaximal knee extensions at 60, 120, 180, 240 and 300 deg/sec with a 60 second rest period between each velocity. All isokinetic tests used a 90^{o} range of motion and included four maximal repetitions.

At the first familiarization session, subjects were fitted on the isokinetic system and settings were recorded to ensure the same positioning for all subsequent familiarization and experimental tests. After the warmup protocol, the subjects performed four maximal contractions at isokinetic velocities of 60, 180, and 300 deg/sec with a 60 second rest between sets. When experimental testing began, subjects were requested to abstain from maximal exercise bouts 24 hours prior to each session and there was a minimum of 24 hours between testing sessions. During the four experimental testing sessions the ascending or descending order of velocities was randomly assigned with each subject performing the ascending and descending order twice at 60, 120, 180, 240, 300 deg/sec with knee flexion held constant at 300 deg/sec. Rest periods between velocity sets were standardized at 60 seconds. Subjects were instructed to contract maximally during knee extension, while flexion velocity was set at 300 deg/sec.

Subjects were given both visual and verbal feedback (VV) during one descending test and one ascending test, while during the corresponding ascending or descending test they received no feedback (NO). When subjects were given feedback, each repetition during the isokinetic set the participant was allowed to view the computer screen which was illustrating effort via line graphs, while verbal feedback consisted of the tester encouraging the participant to “push and pull” though each repetition. The order of feedback was randomized. Each velocity tested was considered a set and the peak torque for each velocity set was used for comparison.

A 2 X 5 repeated measures ANOVA was used to analyze the data with alpha < .05. The independent variables were feedback and velocity sets, as well as, velocity order and velocity sets, while the dependent variable was peak torque.

RESULTS: Analysis of the data revealed no significant differences in peak torque in any of the tests, regardless of feedback (Figure 1 and 2). Additionally, there were no significant differences in peak torque between the corresponding ascending or descending tests regardless of feedback.

DISCUSSION: The primary focus of this study was to evaluate the effect on peak torque when subjects were provided visual and verbal feedback. A second purpose was to assess the differences in peak torque when varying velocity set order (ascending versus descending) during isokinetic testing. Although there were no significant differences in peak torque based on feedback, the mean peak torques of the CA group increased at all velocities with the presence of feedback, which was not true for the RA group. Some studies indicate they have used feedback when the subjects were executing the velocity sets (Dauty et al., 2007; Gioftsidou et al., 2006; Ozcaldiran, 2008;

Tourney-Chollet et al., 2000) but few have compared the results to velocity sets when no feedback has been provided. Considering the manner in which results of velocity sets are reported and how that information is used by clinicians to indicate increases in

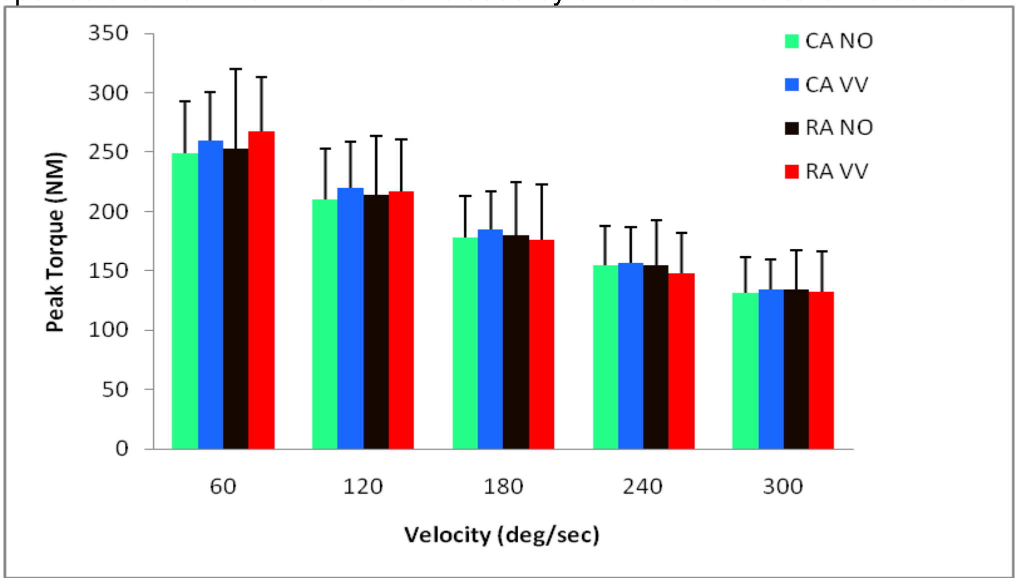

Figure 1. Peak Torques of CA and RA Groups Executing Ascending Velocity Sets With and Without Feedback.

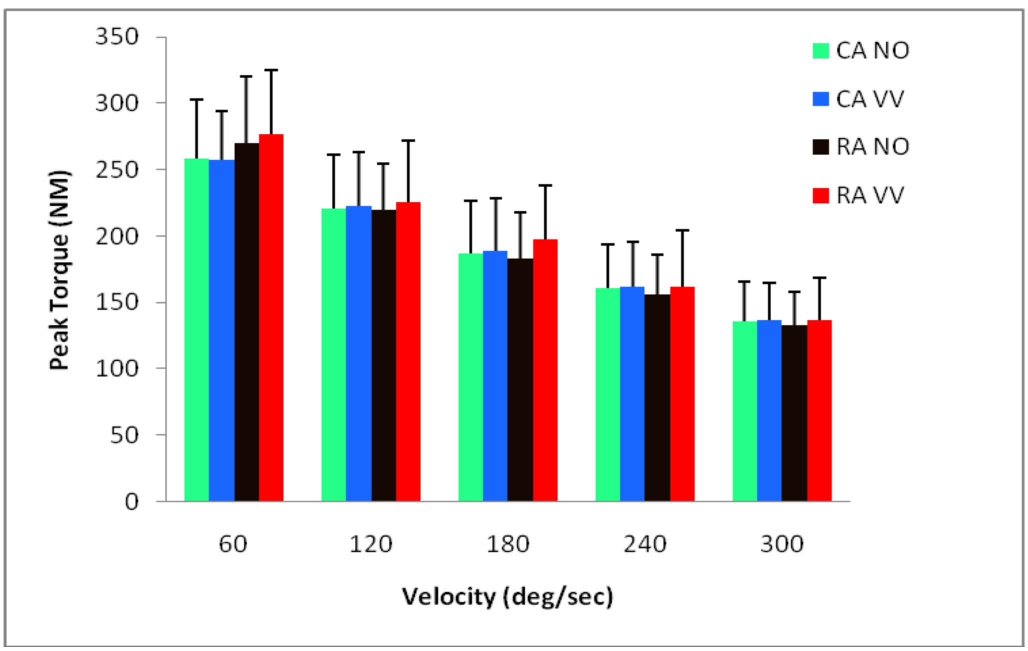

Figure 2. Peak Torques of CA and RA Groups Executing Descending Velocity Sets With and Without Feedback.

strength, rehabilitation progress, etc., it would seem that feedback could be a determinant in a person's effort. Although in this study there were no significant differences in peak torque with the presence or absence of feedback, it would still seem plausible to use feedback to encourage the client during execution.

Although most isokinetic studies seem to use an ascending velocity protocol for assessing peak torque, in the present study the descending protocol produced greater peak torque at all velocities. However, these differences were not significant. Wilhite et al., (1992) suggested that an ascending order of testing should be used when assessing peak torque. Kovelski et al., (1992) and Kovaleski and Heitman (1993) indicated it may be advisable at times to progress from faster velocities to slower velocities. Timm and Fyke (1993) found that order of isokinetic speed did not affect concentric peak torque measurements. In the present study the results seem to point out that a descending protocol would certainly be acceptable for assessing peak torque. From discussions with many of the subjects in this study, it was noted that some favored the descending pattern, while others preferred the ascending order. Therefore, it should be recognized that personal preference of velocity order may play a part in the results of isokinetic testing. If it is the intent of the clinician or researcher to assess the maximum effort of the client/subject, then evaluating which velocity order is most comfortable for the client seems reasonable.

CONCLUSIONS: The conclusions of this study are that peak torque executed during isokinetic sets is not significantly altered by the presence of visual and verbal feedback. Secondly, order of velocity sets maybe a preference of each subject that should be considered when conducting isokinetic tests.

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A KAYAK TRAINING SYSTEM FOR FORCE MEASUREMENT ON-WATER

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KEY WORDS: kayak, force measurement, wireless.

INTRODUCTION: Kayaking is a very competitive sport and represented in the Olympic context with two disciplines: slalom and flatwater. The main forces that propel the boat are paddle and foot stretcher force (Mann & Kearney, 1980). Anecdotal evidence collected from coaches involved in the research suggests varying theories on the best profile and synchronisation of paddle and foot stretcher force. It should be extremely helpful for athletes, coaches and researchers to measure these forces in real-time on-water with an unobtrusive, wireless sensor system such as is presented here. Thereby athletes are provided the possibility to perform their training with knowledge of performance (KP), which leads to superior training effects compared to knowledge of results (KR) only. The authors have not been able to identify any previous studies examining paddle and foot stretcher forces simultaneously although previous work has suggested doing so (Michael et al. 2009, Petrone et al., 1998).

METHOD: The developed system is made up by four units: three waterproof sensor units that link via Bluetooth to a central unit, e.g. a mobile phone, that runs custom software written in Java ME. This central unit accounts for the synchronisation of the data streams and logs performance data with the option to add further functionality in the future. The mechanical loads are sensed with strain gauge (HBM 1-LZ-48-6/350, Darmstadt Germany) setups, which can be screwed onto the paddle, and a modified footrest with four force sensitive resistors (TekScan A201-100, Boston USA) per foot, which replaces the original wooden plate. Each of the three sensor nodes hosts a microcontroller module (EISTEC Mulle V3.2, Luleå Sweden) that converts the analogue signal into 10bit digital data and communicates via, and controls, the Bluetooth radio. Figure 1 visualises the forces that are recorded. The sampling frequency is 150Hz for the paddle sensors and 100Hz for the foot stretcher.

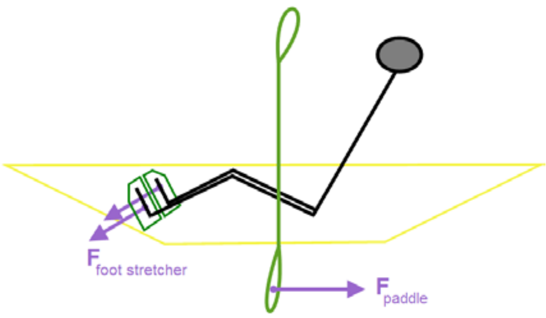

Figure 1. Forces measured by the training system.

Beside the technical aspects usability and compliance with an athlete's expectations were prioritised in the design process. For this reason the sensors can easily be attached to and detached from a kayak requiring no other alteration to the sports equipment. There are no cables between the sensors and they are one-button (power button) operated. A light emitting diode indicates the status of each sensor.

Calibration of the foot stretcher load sensor cells was performed with an electromechanical testing system (Instron 5567, Norwood USA). The foot stretcher unit only has to be calibrated once as it replaces the original wooden plate. The paddle sensors, on the other hand, clamp onto the paddle an athlete chooses to use. They therefore have to be calibrated before every trial by means of loading the paddle with weights at one gripping area while supporting it in the centre and at the other gripping area. The process has to be repeated for the other side.

The system was tested on a kayak ergometer (Dansprint, Hvidovre Denmark) equipped with a rowing computer. Time, total distance, stroke distance, stroke rate and stroke power, amongst other things, can be stored on a personal computer from the ergometer. The ergometer is widely used among Scandinavian elite athletes for indoor training. Non-elite athletes provided data for the evaluation of the system by paddling a distance of 250m.

RESULTS: Force data from each side of the paddle and the footrest were recorded; Figure 2 shows the force profile of a recreational kayak athlete for two strokes, one left and one right. The paddle force profile differs for each side and relates to the associated footrest force. Furthermore, it was found that peak paddle force values and the measured paddle power from the ergometer correlated well (93.6%).

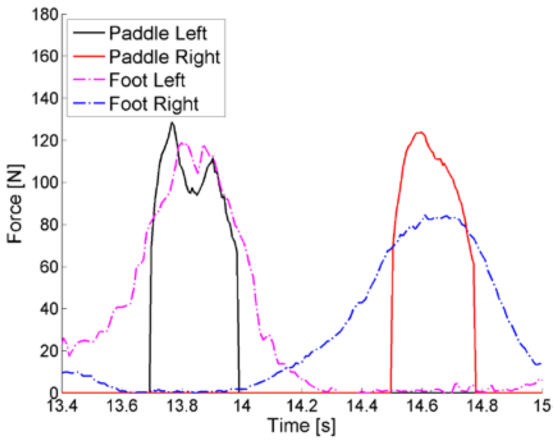

Figure 2. Force profile of two strokes by a recreational kayak athlete.

DISCUSSION: Even though the introduced measuring system has been designed for use on water it was first used on an ergometer because of the machine's possibility to retrieve quantitative data for comparison. The ergometer uses a theoretical model to calculate stroke distance that was not revealed in full to the authors, hence only stroke power data from the ergometer was compared. The results are very promising and motivate the system's use in a study with elite athletes exploring on-water biomechanics from a new perspective.

CONCLUSION: This study presents a novel kayak force measurement system that has been verified in the laboratory and complies with all requirements to be used on-water. The system provides mechanical data with a high temporal resolution for a more thorough analysis and understanding of the kinetics in a kayak.

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THE INFLUENCE OF PRECEDING MOVEMENTS IN THE PERFORMANCE OF BALLET JUMPS

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The purpose of this study was to gain more insights about the individual technique used in the performance of ballet jumps and determine the influence of preceding movements in the efficiency of its performance. Seven female ballet students participate voluntarily in the study. Ground reaction forces were measured using a Bertec force plate. A triaxial accelerometer was placed at the low back of the subject on the skin surface, approximately at the height of the centre of mass. Results suggest that generally, a positive influence from preceding movements was observed in the performance of the selected jumps. Jumps preceded by only one movement, show a tendency to an increase in the reached height. Due to an immature technique of performance, our ballet students did not use appropriately preceding movements to potentiate the subsequent jumps.

KEYWORDS: Ballet, Jumps, SSC.

INTRODUCTION: Classical ballet and modern dance have been shown to be as physically demanding as many team sports (basketball, soccer, football, and volleyball) and very jump-intensive (Nicholas, 1975). Due to the nature of their movements, which involves unusual amplitudes of movement, unusual joint positions, and muscular efforts with excessive impact forces, it is well accepted the significant degeneration of anatomical structures, like bone tissue, tendons and ligaments (Nicholas, 1975; Simpson & Kanter, 1997; Simpson & Pettit, 1997). According to Liederbach et al., (2006), classical ballet dancers perform more than 200 jumps per 1.5-hour daily technique class, more than half of which involve single-leg landing. Despite these, according to some author's the incidence of injuries among dancers is much lower than that among team sport athletes (Orishimo, Kremenic, Pappas, Hagins., & Liederbach, 2009). Several factors unique to dance may be partially responsible for the low injury rates in this population, like technique and intensive training from a young age. Dancers are trained to land with the lower extremities near full extension, with a vertical spine and with maximum use of plantar flexion at initial contact. It is expected that dancers land on the plantar surface of their phalanges and metatarsal heads and immediately roll through their feet, with eccentric control to achieve a quiet heel touchdown and controlling the alignment of the center of the patella directly over the second ray of the foot, trying to dissipate landing forces and to achieve the desired aesthetics of a smooth landing. The goal of the present study was to gain more insights about the individual technique used in ballet jumps, namely the influence that preceding movements can be in the efficiency of its performance. This was done through comparison of kinematical and kinetic parameters in isolated ballet jumps and sequences of movements preceding the same jumps.

METHOD: Seven female ballet students (19.1± 4.1years old, 55.9± kg, 165.0±3.0 cm), with more than ten years of daily practice in classical ballet participated voluntarily in the study. The subjects were fully informed about the purpose, procedures and risks associated with the study and gave their written consent after being informed. The study was approved by the Ethics Committee of the University. None of the dancers present, until the moment, any kind of injury that could influence the performance of the jumps. After a short warm-up, which was individually selected by the ballet dancers, subjects were familiarized with the experimental procedures for data collection. Ground reaction forces were measured using a Bertec force plate (4060-15), with a sample frequency of 1000Hz. Additionally a triaxial accelerometer was placed at the low back of the subject on the skin surface approximately at the height of the centre of mass (Biopac, type TDS109, sensitivity 40 mV/g, range ±50g). Both force-plate and accelerometer signals were sampled at 1000 Hz. After amplifying, all analog signals were converted to digital signals using the 16 bit A/D converter from Biopac. Elementary ballet jumps as well as small sequences of these elementary jumps were selected for the study: Jump 1- squat jump (figure 1 a); Jump 2- counter movement jump (figure 1 b); Jump 3- *Temps levé* (figure 1 c); Jump 4- *Jeté Temps levé* (figure 1 d); Jump 5- *Assemble* (figure 1 e); Jump 6- *Glissade Assemble* (figure 1 e, preceded by a connection element, the small jump *Glissade*); Jump 7- sequence of eight *Sautés* (8* figure 1 b); Jump 8- sequence of six *Jeté Temps levé* (6* figure 1 d); Jump 9- *Entrechat quatre* (figure 1 f); Jump 10- sequence of six *Entrechat quatre* (6* figure 1 f); Jump 11- *Assemble Entrechat quatre* (figure 1 g). The first and second jumps were performed in first position and legs in turn out position. The take-off and the landing phases of jumps 1, 2, 3, 4, 5 and 11 were performed always in the force plate. In Jump 6 only the landing phase was performed on the force plate. Jump 7 was performed in sets of eight consecutive jumps. Jumps 8 and 10 were performed in sets of six consecutive jumps. Three successful trials were required for each jump or sequences of jumps. Jumps were collected in a random order. The determination of jump pairing for subsequent statistical analysis was based on the assumption of being the same jump, or jumps with similar technical requirements, performed in different sequences or isolated. The following parameters were defined for analysis: **t1**- time interval (s) between the beginning of the movement and the moment of losing contact with the force plate; **t2**- time interval (s) between the beginning of new contact with the force plate, after the aerial phase of the jump, and the end of the movement; **f1**- mean vertical force (F_z) as a function of body weight (BW), exerted in the time interval **t1**; **f2**- mean vertical force (F_z) as a function of BW, exerted in the time interval **t2**; **a1**- mean acceleration (m/s/s) in the time interval **t1**; **a2**- mean acceleration (m/s/s) in the time interval **t2**; **t3**- time duration of the aerial phase (s), and **h**- jump height. Values are presented as means and standard deviations (SD) unless otherwise stated. Non-parametric tests were applied for statistical analysis to compare means between the jump pairs. The value of significance was set at α=0.05.

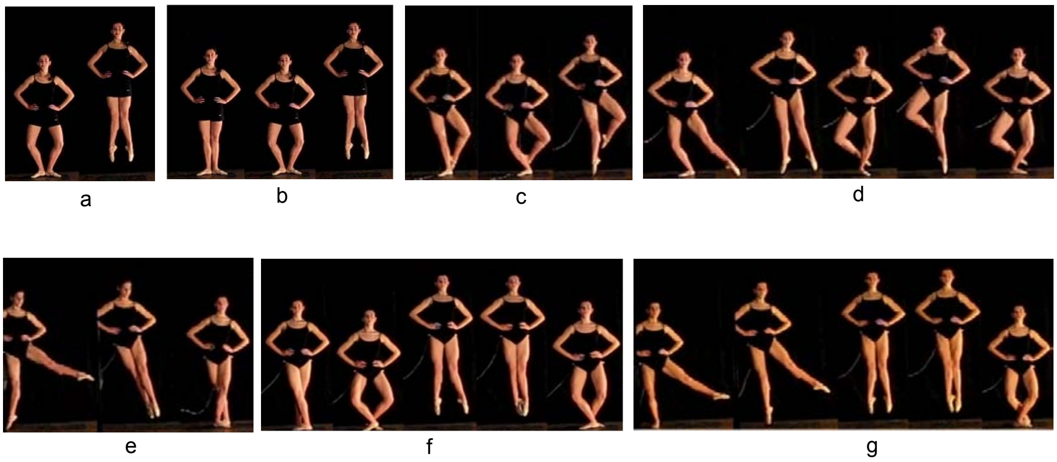

Figure 1. Elementary ballet jumps selected for the study: a) squat jump; b) counter movement jump; c) *Temps levé*; d) *Jeté Temps levé*; e) *Assemble*; f) *Entrechat quatre*; g) *Assemble Entrechat quatre*.

RESULTS: Table 1 presents the average values (±SD) of all the parameters selected for analysis in each jump.

Table 1. Average and SD values for time duration (t1 and t2), F_z force (f1 and f2), acceleration (a1and a2), time duration of the aerial phase (t3), and jump height (h) in all jumps and in all subjects.

Jump	t1 (s)	t2 (s)	F1 (F/BW)	F2 (F/BW)	a1 (m/s/s)	a2 (m/s/s)	t3 (s)	H (cm)
1	0,13±0,02	0,10±0,00	2,62±0,20	2,88±0,46	1,52±0,28	1,79±0,60	0,35±0,00	15.44±1.68
2	0,16±0,02	0,10±0,01	2,70±0,21	2,83±0,32	1,51±0,34	1,58±0,45	0,38±0,02	18.01±2.07
3	0,18±0,14	0,13±0,00	2,05±0,17	2,52±0,25	0,79±0,17	1,47±0,34	0,23±0,02	6.51±1.10
4	0,21±0,47	0,13±0,01	2,43±0,35	2,63±0,25	1,15±0,24	1,50±0,26	0,23±0,04	6.81±1.83
5		0,11±0,02		2,74±0,39		1,77±0,56		
6		0,10±0,01		2,88±0,45		1,81±0,70		
7	0,18±0,02	0,09±0,01	3,48±0,38	3,12±0,45	2,51±0,62	2,42±0,75	0,36±0,03	15.70±2.48
8	0,21±0,03	0,12±0,01	2,66±0,03	2,34±0,21	1,56±0,24	1,25±0,38	0,23±0,02	6.40±1.12
9	0,14±0,01	0,10±0,00	2,94±0,26	3,06±0,26	1,83±0,41	1,92±0,40	0,37±0,02	16.59±1.89
10	0,17±0,02	0,09±0,01	3,58±0,53	3,14±0,39	2,58±0,42	2,22±0,52	0,37±0,03	16.58±2.85
11	0,13±0,01	0,09±0,01	3,36±0,53	3,06±0,36	2,30±0,46	2,02±0,48	0,38±0,04	17.56±3.16

Table 2 presents the p-values for the Wilcoxon test between the jump pairs. We paired jumps 1 and 2; jumps 2 and 7; jumps 2 and 5; jumps 3 and 4; jumps 3 and 8; jumps 4 and 8; jumps 5 and 6; jumps 9 and 11; jumps 9 and 10; and finally jumps 10 and 11.

Table 2. p-values for the Wilcoxon test between jump pairs in time duration (t1 and t2), F_z force (f1 and f2), acceleration (a1and a2), time duration of the aerial phase (t3), and jump height (h).

Jump Pairs	t1	t2	F1	F2	a1	a2	t3	h
1- 2	0,028	0,173	0,345	0,753	0,917	0,463	0,028	0.028
2- 7	0,173	0,028	0,028	0,116	0,046	0,028	0,028	0.028
2- 5		0,917		0,753		0,249		
3- 4	0,173	0,249	0,046	0,345	0,028	0,753	0,917	0.917
3- 8	0,116	0,249	0,028	0,249	0,028	0,917	0,753	0.753
4- 8	0,173	0,463	0,116	0,345	0,075	0,249	0,345	0.345
5- 6		0,075		0,173		0,753		
9- 11	0,115	0,080	0,116	0,917	0,116	0,093	0,173	0.173
9- 10	0,116	0,046	0,046	0,600	0,046	0,075	0,917	0.917
10- 11	0,028	0,141	0,116	0,917	0,046	0,463	0,075	0.075

DISCUSSION: As it possible observe in table 1, the time duration of the aerial phase was significantly higher in jump 2 (counter-movement jump in first position and legs in turn out position), and in this way, the maximal vertical displacement was also observed in this jump. These specific particularities lead us to the concept of stretch-shortening cycle (SSC) (Norman & Komi, 1979). The forced lengthening of an active muscle before allowing it to shorten, leads to an enhanced response during the shortening phase. This potentiation phenomenon is a very common muscle action present in many human movements. In the same way as in the performance of the sportive jumps, ballet jumps follow the same potentiation mechanism of the SSC. Unfortunately, due to the deficient technique used by these ballet students in his performances, the potentiation phenomenon was not always registered in all jumps. Higher values of force (F_z) as a function of BW, in both phases of impulsion and reception, were observed in Jump 10 (a sequence of six *Entrechat quatre*). This jump is a highly demanding jump belonging to the *batterie* exercises in a ballet lesson. During the aerial phase of these jumps, dancers should perform a fast and precise exchange of legs with a controlled and smooth landing. It is also possible to observe that, as expected, in jumps previously preceded by the *Jeté* action (Jumps 4 and 8), the time duration **t1** (time interval between the beginning of the movement and the moment of losing contact with the force plate) was higher. This action involves a sliding of one foot in the ground, during a *plié* movement, to increase the contact time and extend the impulsion phase in order to improve the amplitude and efficiency of the following jump. Unfortunately, this complex action was not well used by our students in the connection to the following jump, reflecting a losing of potentiation and probably of the total energy. Accordingly, this fact can be attenuated with a proper technique and intensive training programs from a young age (Orishimo et al., 2009). These facts can also be observed in table 2. The major number of parameters significantly related was only observed in the jump pairing between jumps 2 and 7. Contrary to the expected, in the other jump pairs the number of parameters significantly different was less marked.

CONCLUSION: Results suggest that, generally, a positive influence from preceding movements was observed in the performance of the selected jumps. Jumps preceded by only one movement, show a tendency to an increase of the reached height. Due to an immature technique of performance, our ballet students, especially in the sequences of small jumps, did not use previous movements to potentiate the following jump tasks. More research should be done in this area and future studies should include more subjects for analysis and with a better technique of jump performance.

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SAGITTAL PLANE RESISTANCE TORQUE IN ANKLE BRACES

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KEYWORDS: ankle brace, angular torque, ankle joint

INTRODUCTION: Ligaments of the ankle joint complex are among the most frequently damaged structures during sports and physical activity (Eils et al., 2002). One common intervention used to prevent ankle ligament injury is the application of lace-up style ankle braces. These braces, usually made of non stretch nylon materials, increase the mechanical stability at the ankle joint by restricting the allowable range of motion thereby limiting strain on joint ligaments. Ankle braces are primarily designed to restrict motion in the frontal plane to limit ankle inversion and eversion without impeding the plantar-dorsi flexion (PF) motion (Eils et al., 2002). However, studies examining the effect of bracing on ankle motion during drop jumping have found a significant reduction in sagittal plane ankle motion while braced (DiStefano et al., 2008). Previous studies have examined isolated ankle range of motion restriction around the PF axis with different brace types (e.g. Eils et al 2002), but these studies were not able to distinguish the resistance torque due to the brace alone. The purpose of the present study was to measure the passive mechanical resistance torque around the ankle PF axis generated by a range of commercially available ankle braces while moving through the sagittal plane.

METHODS: Five widely used commercial ankle braces were examined (Table 1). All models were a lace-up design. One brace (ASO) was tested with and without removable plastic lateral supports (stays). A mechanical shank/ankle/foot was used to simulate passive motion around the PF axis. The shank was a sculpted wood blank and the foot was a 26 cm prosthetic foot. The shank and foot were connected by a mechanical ankle composed of a single revolute hinge simulating the PF axis. The PF axis was aligned to be perpendicular to the sagittal plane. The mechanical shank was rigidly attached to an isokinetic dynamometer (Humac Norm, Computer Sports Medicine, Inc., MA) with the PF axis of the ankle aligned with the axis of rotation of the dynamometer (Figure 1).

Table 1. Five brace designs tested. The ASO brace was also tested without plastic lateral supports.

Ankle Lok (Swede-O, North Branch, MN)
ASO (Medical Specialties, Inc., Charlotte, NC)
ASO without stays (Medical Specialties, Inc., Charlotte, NC)
Pro Lacer (Cramer Products, Inc.,Gardner KS)
AS1 (Cramer Products, Inc.,Gardner KS)
ATF (Mueller Sports Medicine, Inc., Prairie du Sac, WI)

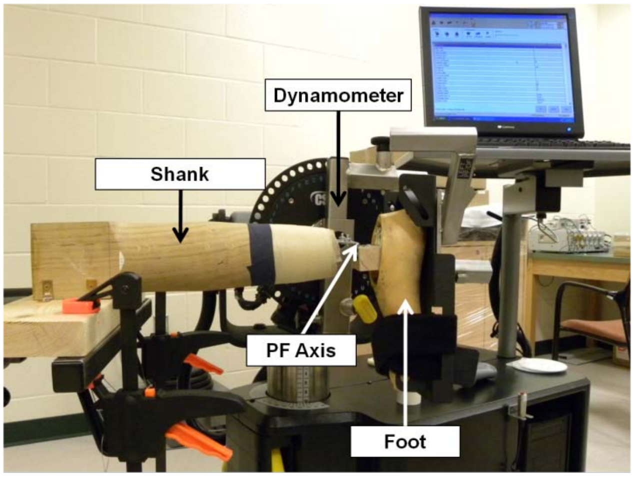

Figure 1. Mechanical shank/ankle/foot attached to dynamometer

Braces were applied to the mechanical shank/ankle/foot according to manufacturer's directions. Braces were tightened in a similar manner and initial tightness was controlled using a thin-film pressure transducer (FlexiForce, Tekscan, Inc, MA) placed under the laces in the proximal section of each brace. The dynamometer cycled each brace through a physiological range of motion (30° dorsiflexion to 60° plantar flexion) while simultaneously recording the resistive torque in the PF axis. All braces were cycled 10 times at a constant speed of 10°/s and torque vs PF angle profiles were obtained. A reference trial which no brace was attached to the shank/ankle/foot was obtained to account for residual torque from the experimental set-up and gravity effects. This residual torque was accounted for when calculating brace torques.

RESULTS: The torque produced by each brace differed over the entire range of motion with the greatest resistance torque produced during maximal plantar flexion (Figure 2). The ASO and AS1 ankle braces generated the largest torques (Figure 2). The neutral position (i.e. where no resistance torque was found) ranged between 12° dorsiflexion to 5° plantar flexion (Figure 2). The ASO brace tested with the plastic lateral supports removed showed a distinct decrease in PF resistance torque as well as an altered neutral position compared to the complete ASO with lateral support stays. Although all braces were measured at a velocity of 10°/s, different speed trials were performed (75°/s and 180°/s) with no difference in torque outputs and as such are not reported.

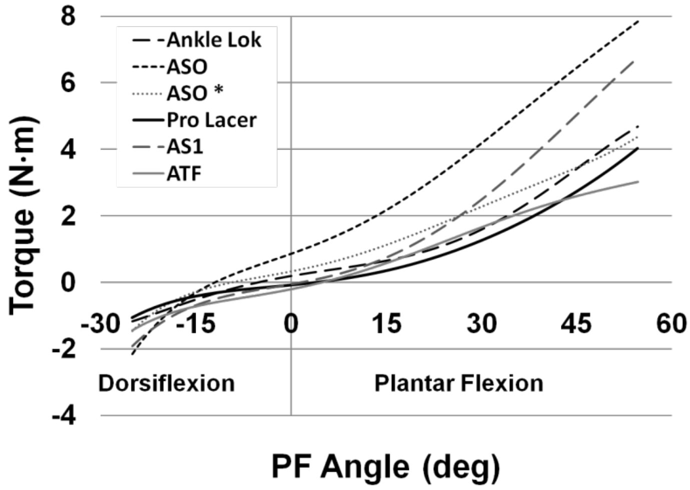

Figure 2. PF torque vs angle. (ASO* = without lateral supports)

DISCUSSION: The primary goal of this study was to determine the passive mechanical resistance torque in PF axis generated by 5 different commercially available ankle braces. The PF resistance torque was proportional to PF angle and there appeared to be differences between commercial brace designs. These observed differences in torque can be attributed to the material composition of each brace as well as subtle differences in lacing and support strap designs. The lateral plastic supports in the ASO brace markedly increased the passive PF resistance torque indicating a coupling between frontal and sagittal brace stiffness.

CONCLUSION: This study identified the passive mechanical resistance torque around the PF axis in a series of lace-up ankle braces generated during sagittal plane motion. Although ankle brace designs aim to limit the frontal plane of motion at the ankle, our results indicate that sagittal plane torque also occurs. This information will be useful when modeling ankle joint motion while wearing braces. In particular, when performing inverse dynamics analysis this type of information is needed to separate torques due to muscle and soft tissue action from those due to the passive mechanical properties of the brace alone.

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COORDINATION DURING INITIAL ACQUISITION OF THREE-BALL JUGGLING

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KEYWORDS: Complexity, elbow, information

INTRODUCTION: Bimanual coordination is critical for the performance of a number of continuous skills (e.g., walking, running, drumming, etc.). However, relatively little is known about how such coordination is developed and maintained. In the current study researchers sought to address this issue by examining coordination changes in bilateral elbow motion during initial acquisition of three-ball cascade juggling. Elbow motion was assessed using a set of both traditional (Average Amplitude; *Avg. Amp*) and more recently developed *nonlinear* time-series analyses (Approximate Entropy; *ApEn* – a measure of *complexity* within a single time series, and Average Mutual Information; *AMI* – the amount of information *shared* between two time-series) (Abarbanel, 1996; Pincus, 1995). It was hoped that together these analyses would lend new insights about bimanual coordination in continuous skills that could be used to develop more sophisticated biomechanical models of human movement and/or advance general motor theory

METHODS: Eighteen college-aged students underwent twelve supervised training sessions (three sessions per week for four weeks, each session lasting approx. 45 minutes) during which participants received juggling instructions provided by experimenters (Finnigan et al., 2002). Once able to achieve ten consecutive tosses participants began to undergo three experimental trials at the end of the next nine training sessions where elbow motion (relative elbow flexion in degrees) was recorded at 140 Hz for the fist 7-sec of each trial where participants were instructed to complete ‘as many juggling tosses in a row as possible’. Elbow motion was recorded from electronic sensors placed bilaterally on the upper arms (midpoint between acromion process and cubital fossa) and forearms (midpoint between cubital fossa and ulnar styloid) using a magnetic tracking system (Flock of Birds, Ascension, Inc.) interfaced and digitized with MotionMonitor Software® (v. 7.72). All elbow data were then analyzed with *ApEn, Avg. Amp*, and *AMI* using custom Matlab code. All dependent measures (including number of successive tosses) were averaged across all three experimental trials for each training session where elbow data were recorded.

RESULTS AND DISCUSSION: Participants saw significant improvements in juggling performance with training. $F(8,136)=11.36$, $p<.01$ (Fig.1).

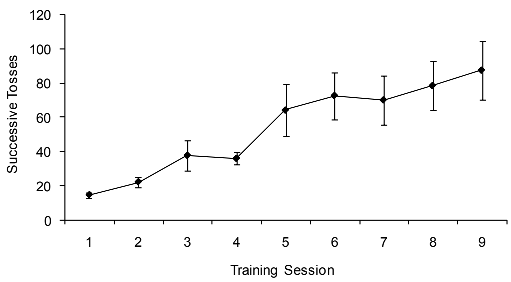

Figure 1. Mean ± standard error (SE) tosses completed in a row during experimental trials

Participants also showed a significant increase in *ApEn*, $F(8,272)=3.98$, $p<0.01$, accompanied by a decrease in *Avg. Amp*., $F(8,272)=7.17$, $p<0.01$ (collapsed across dominant/non-dominant limbs) (Fig. 2). These findings indicate that as training progressed elbow motion became confined to an increasingly smaller range (as indicated by a steady decrease in *Avg. Amp*), but the movement pattern within that range became more *complex* (as indicated by an increase in *ApEn*).

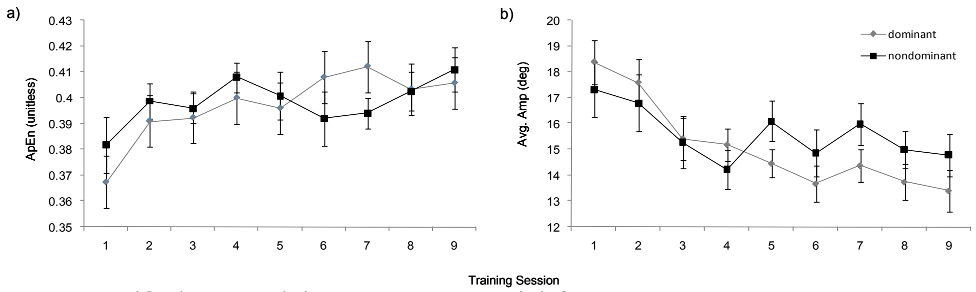

Figure 2. Mean (SE) *ApEn* (a) and *Avg. Amp* (b) for successive training days

Finally, participants showed a progressive decrease in *AMI* with training, *F*(8,136)=6.43, p<0.01 (Fig. 3). This finding indicates that the amount of information *shared* between two limbs was steadily decreasing. This was interpreted to reflect decreased congruence in movements of the two limbs.

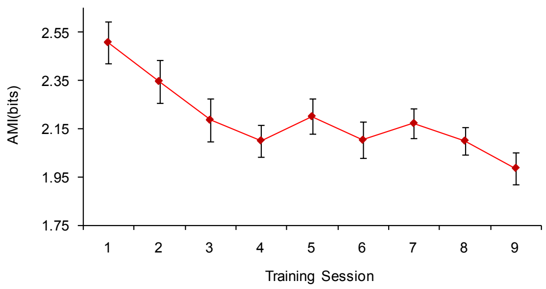

Figure 3. Mean (SE) AMI values for successive training days

CONCLUSION: The analyses used in this study successfully revealed a number of unique features in bimanual coordination during initial acquisition of three-ball juggling. For intra-limb coordination results showed that elbow motion became more confined, but also more *complex*. Might this reflect the steady development of a coordination that exhibits flexibility in one dimension (movement pattern) and stability on another (range of motion)? Assessment of inter-limb coordination showed that movements of dominant and non-dominant limbs were had become less congruent with training. This finding might indicate that control over the two limbs was progressively becoming more independent and could be hinting at the possibility that each limb has a different role to play in the coordination of this skill? In anecdotal observations experimenters noted that towards the end of the training some participants seemed to be using their dominant limb to set the tempo for juggling and the non-dominant limb to make adjustment and corrections for inconsistencies with each toss. At present more work is needed to replicate these findings and investigate possibility that different limbs may have independent roles to play in bimanual coordination in juggling as well as other bimanual skills (e.g., running and walking gait).

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QUANTIFICATION OF TIME TO STABILISATION USING THE SEQUENTIAL ESTIMATION TECHNIQUE

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KEYWORDS: time to stabilisation, dynamic balance, ground reaction force

INTRODUCTION: Dynamic postural stability is a key physical attribute for athletes and can be defined as the ability to maintain postural balance whilst moving from a dynamic to a static state (Wikstrom et al., 2005). This dynamic stability requires the complex integration of sensory afferent systems (visual, somatosensory and vestibular) with efferent responses in both upper and lower body neuromuscular systems. The effect of neuromuscular pathologies on dynamic postural stability has been the focus of a considerable body of research (e.g. Ross et al. 2009), particularly in the areas of chronic ankle instability and ACL injury. Underpinning this research are several functional tests that have been proposed to objectively quantify dynamic postural stability. One of the most commonly cited measures from these tests is known as 'time to stabilisation' (TTS) which is defined as the time required to reach stability after landing. As reaching stability is a somewhat non-specific event, several different techniques have been proposed for quantifying the TTS. The majority of techniques use a generalised approach that identifies when the resultant ground reaction forces (GRFs) reach some baseline threshold. One such TTS technique was proposed by Colby et al. (1999) and has since been used by numerous others (e.g. Shaw et al. 2008). This technique uses a process called sequential estimation, whereby a cumulative average of GRF data is calculated by adding one data point at a time. Stabilisation is deemed to occur when the cumulative average reaches and stays within 0.25 SDs of the overall mean. This sequential estimation procedure has been applied to mediolateral (M-L), anterior-posterior (A-P) and vertical GRF data from a range of jump types. While data from this technique has not always supported expected group or condition differences in stability, the technique is still well recognised and used in TTS calculation.

To the author's knowledge, no research has pointed to mechanistic flaws in the use of sequential estimation in calculating TTS values. However, on reviewing the technique, it appears that the use of both the cumulative average and SD values predispose the technique to inaccurate TTS assessments. Use of a cumulative average suggests that increased force oscillations can theoretically reduce the TTS, and similarly, the use of SD values based on the full landing sequence implies that the threshold range is larger for less stable landings and can thus also theoretically lead to shorter TTS values. In seeking to understand the links between dynamic stability and neuromuscular pathologies, it is clearly essential that the measures used are robust and do not provide spurious results. Therefore, the aim of this paper was to assess the validity concerns of the sequential estimation method of TTS calculation by using it to compare jumps with clear differences in dynamic stability.

METHOD: A healthy male subject (age 29 years, mass 89.5 kg, leg length 95.5 cm) completed 10 hop trials in each of two conditions. In condition one (stable), their goal was to stabilise as quickly as possible after landing and hold the stable posture for a further three seconds. In condition two (unstable), they were asked not to stabilise immediately on landing but to continue swaying in the A-P plane on one leg. Condition two was used to simulate poor dynamic stability. The hops were conducted in accordance with the protocol used by Colby et al. (1999), and in brief involved a single-legged forwards hop and landing onto the force plate from a distance of one leg length. Data from a piezoelectric force plate (Kistler AG, Winterthur, Switzerland) were recorded in A-P, M-L and vertical planes at 1000 Hz for 3 seconds after impact (force > 5 N). TTS values were calculated using the sequential estimation approach presented by Colby et al. (1999) for GRF data in all three planes for all

20 hops. Differences between TTS scores for each condition were assessed for statistical significance using an independent samples t-test in SPSS (v.17) with an alpha level of 0.05.

RESULTS AND DISCUSSION: Qualitatively, clear differences were evident between force traces for unstable and stable landings (see Figure 1), which supported the correct execution of both landing conditions. However, vertical force TTS values were significantly shorter (p=.005) in unstable landings than stable landings (see table 1 for mean TTS scores). This indicates particularly poor validity for this method of TTS assessment as unstable landings should actually show longer TTS values. No significant differences were shown for A-P and M-L TTS values between the conditions (p=0.30 and 0.20, respectively). The sequential estimation technique may hold an advantage over other methods of TTS assessment as its protocol is clearly documented and can be carried out easily in MS Excel, however, the data from this study shows that it cannot even differentiate correctly between extreme conditions of stable and unstable landings, and in fact portrays the unstable landings as being the most stable when considering TTS of vertical GRFs.

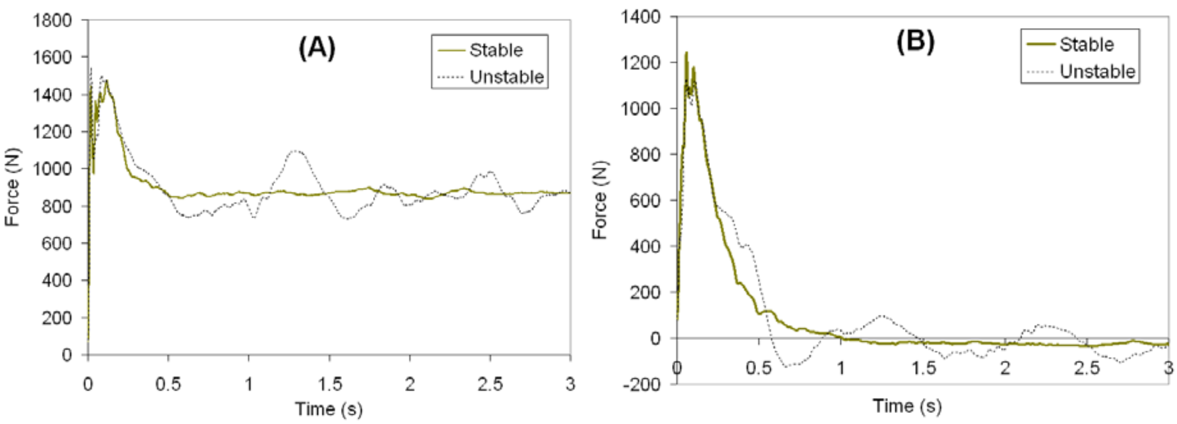

Figure 1. (A) Exemplar vertical force traces for stable and unstable landings, where stable TTS = 1.44s and unstable TTS = 1.03s. (B) Exemplar A-P force traces for stable and unstable landings, where stable TTS = 1.80s and unstable TTS = 1.78s.

Table 1. TTS scores (mean±SD) for stable and unstable landings using sequential estimation

	Vertical TTS (s)	Anterior-Posterior TTS (s)	Mediolateral TTS (s)
Stable	1.44 ± 0.03	1.79 ± 0.02	0.58 ± 0.44
Unstable	1.18 ± 0.17	1.80 ± 0.04	0.88 ± 0.54

CONCLUSION: The results of this study question the validity of the sequential estimation technique for calculating TTS with A-P, M-L or vertical GRF data. Practitioners and researchers with an interest in dynamic stability should note the presented limitations of this technique when reviewing literature or planning future studies in the area.

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BELAY TECHNIQUES ON STOP FALLING OF A CLIMBER

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The purpose of the present study was to identify the kinematic differences between two popular rock climbing belay techniques used in the United States: brake, under, slide (BUS) and slip, slap, slide (SSS) and their efficiency in stopping the fall of a climber. Five male subjects with no previous belaying experience from two different beginning rock climbing classes were recruited to participate in the study (n = 10). Each subject passed a practical belay test before participating in the study. Data of time to stop the fall of a climber, vertical displacement of the falling climber, maximum negative vertical velocity of the falling climber, and percentage of time each belayer was in the braked position were collected. Results showed that the climber had significant greater falling displacement, longer time to stop, and higher maximum negative velocity when SSS was performed.

KEYWORDS: rock climbing, belay technique, BUS, SSS.

INTRODUCTION: The global popularity of rock climbing has soared as it has evolved from a competitive sport in extreme outdoor landscapes to trendy recreational activity with indoor climbing gyms (Long, 2003). Over the last decade in the US alone, the number of individuals involved in rock climbing is rapidly approaching nine million individuals (Outdoor Industry Foundation, 2006). The research has surged in this area, with the biomechanical literature in particular, concentrating mostly on the climber's physiology exertion, technique of climbing, and climbing equipment such as belay device and dynamic rope (i.e., Bourdin, Teasdale, Nougier, Bard, & Fleury, 1999; Noẻ, Quaine, & Martin, 2001; Quaine & Martin, 1999; Vogwell & Minguez, 2007).

Since rock climbing presents with natural risks and is drawing an ever diverse population of individuals ranging both in age and experience, professionals in the field are constantly vigilant for new technology and techniques to reduce the possibility of injury. According to Nelson and McKenzie (2009), tens of thousands of individuals were hurt over a span of two decades with falls accounting for over 75% of total injuries. Falling, of course, cannot be eliminated from climbing, it is an accepted risk; however, the manner in which a fall is safeguarded can be managed and changed to reduce the risk of injury. Belaying is a method used to slow and arrest a fall in climbing. In a climbing situation, a rope is tied to the climber and to the person belaying – the belayer. A safe belay incorporates an anchor to hold the belayer in place, a body position that anticipates the direction of force applied to the belayer, good communication and the climber and belayer, and friction applied to the moving to slow and stop the falling climber (Powers, 2009).

Several aspects of belaying may be able to neutralize the negative consequences of a climber's fall: 1) the dependability of the equipment, 2) the belayer's skill level, and 3) the belay technique. Due to technological advances in equipment, the first variable can be considered to be constant. Differences within any particular skill level can also be stabilized through instruction, practice, and proficiency testing. The two common rock climbing belaying techniques are the brake, under, slide (BUS) and the slip, slap, slide (SSS). Both methods of belaying are very similar in that the slack is managed with the use of the hands and the belay device; however, they differ in the movements of the belayer's hands after taking in the slack. The SSS technique requires moving the brake hand with the rope upwards to meet the guide hand. The guide hand then "slaps" the two portions of rope together to allow the brake hand to slide along the rope toward the belay device (Stiehl & Ramsey, 2005). In contrast, the BUS method involves the guide hand letting go of the rope and moving underneath the brake hand so that the brake hand can slide along the rope toward the belay device (Stiehl & Ramsey, 2005). In both methods, the brake hand never lets go of the rope. The BUS technique, in theory, provides longer duration of braking phase of the rope when compared to the SSS technique.

Although there is no evidence that connects climbing injuries to the type of belay method, some climbers and some organizations teach one method over the other claiming that it is safer. A search through Accidents in North American Mountaineering for the past ten years shows no evidence of climbing accidents that can be attributed to the type of belay method used (Williamson, 2009). Accidents have been ascribed to failed belay anchors, or improper belaying techniques, but there is no empirical evidence to suggest that either the SSS of the BUS is a superior method of belaying (Williamson, 2009; Nelson & McKenzie, 2009). According to impulse-momentum relationship and gravity, the greater falling distance of a climber, the greater changing velocity the climber has. Therefore, if one method can arrest a fall faster, thus decreasing the distance of a fall the chances of injury would certainly be minimized (Nelson & McKenzie, 2009) such as injury from hitting a ledge. In addition, the dynamic rope has its mechanical property to elongate due the force applied. It is important to minimize the fall distance and duration within the dynamic rope's safe limits. Therefore, the purpose of the present study was to identify the kinematic differences between two belay techniques: BUS and SSS and their efficiency to stop the falling of a climber.

METHOD: Ten male subjects were recruited from two different rock climbing classes. Five participants were from a climbing class that taught the BUS and the other half were from a class that taught the SSS. All of the subjects received informed consent and all policies and procedures of using human subjects was followed and approved by the Human Subject Review committee at the local university. None of the subjects had any previous rock climbing or belay instruction. One month of instruction (one 2-hour class per week) in the university climbing gymnasium), and practice were given to learn either the BUS or SSS belay techniques. All subjects passed a practical belay exam given by their instructors before they could participate in the study.

Before data collection, all subjects were instructed to perform the technique in the plan of motion (POM) and markers were placed at the following joint axis on the right side of the body: shoulder, elbow, wrist, hip, knee and ankle. All subjects performed the belay techniques with the same climber, on the same climb using the same rope and belay device. In order to determine the fall of the climber, one marker was placed on the 4^{th} lumbar vertebra and the climber was instructed to climb straight up and fall any time after point A and before point B (1 meter from the top of the camera view) (Figure 1). The belay performer was not informed that when and where the climber would fall.

One 60 Hz high speed camera (Panasonic, model: 5100 HS) was set up 10 meters away from the POM and 1.63 meters above the floor. Each subject was required to perform three successful belay trials using the technique on which he had been trained. All successful trials included the belay performer staying in the POM and performing the proper technique during the climber's fall. The video was analyzed from the 10^{th} frame before the falling of the climber to the 10^{th} frame after the lowest falling position of the climber.

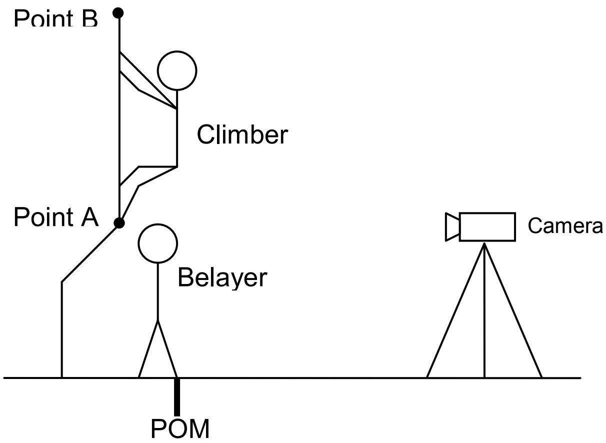

Figure 1. Experiemental setup (Note: Figure not drawn to scale).

Vertical displacement of falling includes the free fall period and the vertical displacement due to the elastic property of the dynamic rope. Time of falling was calculated from the beginning of the vertical downward movement to the lowest position of the fall. Time for belayer to reach the locking position (e.g., the angle between the dynamic rope on either side of the belay device is equal to or greater than 90 degrees) when he first sees the falling, which includes reaction time and motion time. Multiple t-tests were performed to examine the statistic significance between two types of belay techniques. In order to avoid type I and II error, the level of significance was readjusted according to the number of the comparisons using Holm's correction (*P*-value $= \alpha/(n-i+1)$), where *n* is the total number of comparisons and *i* is the order of comparison (Knudson, 2009; Lundbrook, 1998).

RESULTS: Results indicated that the climber in SSS belay technique group had significant greater vertical displacement, maximum velocity, and a longer falling period with *P*-value less than the new adjusted *P*-value of 0.05. However, due to some discrepancies in the belayers' hand positions at the time of the fall, the percentage of the time in the braked position could not be measured. In addition, the performance in arresting the fall had large variations from trial to trial which resulted in the difficulty of identifying the beginning and the end of belay performance to catch the falling climber. Though the cycles of pulling slack from both techniques are different, belayers failed to complete a whole cycle of pulling slack of dynamic rope during the fall and in some trials, belayers started a new cycle of pulling slack. In other trials, the belayer assumed the locking position after finishing only part of the pulling cycle for both techniques. Therefore, this issue of pulling cycle resulted in the difficulty to compare times for the belayer to get into lock position. Due to the timing of getting into lock position varied from each individual belayer, the displacement and the time of the falling climber were also varied.

Table 1. Kinematic variables of the climber

Techniques	Displacement (m) *	Maximum velocity (m/s) *	Time to stop (s) *
BUS	.73 ± .04	2.17 ± .12	.51 ± .02
SSS	.90 ± .10	2.61 ± .44	.57 ± .01

Note: An * indicates significant difference with newly adjusted *P*-value.

DISCUSSION: The use of belaying techniques is an essential safety feature for rock climbers, especially for beginners. Therefore, two frequently utilized belaying techniques, BUS and SSS, were compared in terms of the efficacy of arresting a climber's fall. The main objective of belaying is to arrest a climber's fall quickly with as little vertical distance and velocity as possible. When the fall distance increases, the fall velocity, of course, will increase due to gravity before the elastic component of the dynamic rope starts to elongate and transfer the kinetic energy to strain energy and arrest the fall. One notable study showed that one of the most common injuries from rock climbing is back strain which is due to a sudden stop from the suspension of the rope (Hohlrieder, Lutz, Schubert, Eschertzhuber, & Mair, 2007). Although with better technology, the dynamic rope can be lengthened to absorb the shock from a sudden stop, the greater displacement of a fall still increases the risk of injury.

For novice belayers, the BUS technique appeared to be more effective in reducing the amount of displacement. In theory, belay performance can be separated into two phases of braking and pulling slack. As the climber moves upward, there is a period when the brake hand has to be repositioned while the belayer is taking in slack. This is a vulnerable period since the dynamic rope is not in a braked position. When the BUS technique applied, this vulnerable phase is shortened and the rope remains in a braked position for a longer duration. Therefore, it is more advantageous for the belayer to get into the braking position when using the BUS technique, which may have accounted for the findings that BUS has shorter amount of time to stop the falling in the present study.

Since there were difficulties to examine the time to assume locking position, it is hard to explain if BUS technique has less range of motion during a cycle so which resulted in higher efficiency when pulling the slack of the dynamic rope as the climber is ascending. Therefore, in order to analyze the belay techniques kinematically, the suggestions for future study are: 1) capturing three cycles of pulling slack of the rock until the locking position occurred, 2) using skilled belayers for both techniques to see if the same results of falling occur, and 3) separate the falling distance into free fall and the displacement due to elastic component of the dynamic rope.

CONCLUSION: The purpose of the present study was to investigate two common belay techniques (BUS and SSS) that are used to catch the falling of a climber. The findings suggested that after one month of instruction for all the beginners, the BUS group showed greater efficiency of catching the fall of a climber with a shorter period of time to arrest the climber with less vertical displacement and smaller maximum vertical velocity of the climber's fall when compared to SSS technique. Therefore, the BUS technique is suggested for beginners at the entry level since it is more efficient to stop the fall of a climber. However, this may not apply to skilled belayers who may be used to performing either technique.

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RACE PATTERN OF 60-M HURDLES IN WORLD-CLASS SPRINT HURDLERS: A BIOMECHANICAL ANALYSIS OF WORLD INDOOR CHAMPIONSHIPS 2010

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KEYWORDS: Competition, Kinematics, Athletics

INTRODUCTION: Earlier competition reports of sprint hurdle performances show that the world-class hurdle sprinters achieve their maximal race velocity somewhere between the third and sixth hurdle (Brüggemann & Glad, 1990; Brüggemann et al., 1999). No competition analysis has been done for the indoor sprint hurdle performance, where the official distance is 60 meters consisting of five hurdles. It is not known if the hurdle sprinters change their race pattern towards a more aggressive start in the 60-m hurdle race. The purpose of the present study, therefore, was to examine the race pattern of 60-m hurdles in world-class sprint hurdlers.

METHOD: The data was collected during the IAAF World Indoor Championships in Athletics (Mar 12-14, 2010, Doha, Qatar). The finals of 60-m hurdle races were filmed by two high-speed video cameras (300 fps) set above the spectator stands close to hurdles three (H3) and five (H5). The cameras were synchronized by the light signal of the starting gun and panned to follow the athletes throughout the race.

The instants of take-off (TO) and touchdown (TD) to each hurdle were analyzed from the video recordings. Intermediate times (H1-5) were calculated from the start (gun signal) to each hurdle TD. Interval times (H1/H2, etc) represent the time between two consecutive hurdle TDs. Hurdle clearance times were calculated from TO to TD for each hurdle. Run-in times were calculated from the final results minus the 5th intermediate times. Final results were provided by SEIKO (official timing for the event). Pearson's correlation coefficient was used to examine the relationships between selected variables. Level of significance was set at P<0.05.

RESULTS: The main data is presented in Table 1. All the men sprint hurdlers achieved their fastest interval by H4, whereas three of the women finalists got their fastest interval only between H4 and H5. The maximum interval velocity (MaxInterV) for men and women is 9.12 ± 0.19 ms^{-1} and 8.72 ± 0.18 ms^{-1} respectively. The fastest interval times were significantly correlated to race result both in men and women (r = -.74, P<.05 and r = -.83, P<.05, respectively).

Table 1. Time analysis of 60-m hurdle finalists in World Indoor Championship 2010. (mean±SD)

	Time[s]											
	H1	H2	H3	H4	H5	Result	H1/H2	H2/H3	H3/H4	H4/H5	Run-in	Mean ClearT
Men (n=8)	2.57 ±0.08	3.61 ±0.08	4.62 ±0.11	5.63 ±0.14	6.66 ±0.16	7.53 ±0.16	1.04 ±0.02	1.02 ±0.03	1.01 ±0.03	1.02 ±0.03	0.88 ±0.01	0.33 ±0.02
Women (n=8)	2.61 ±0.04	3.68 ±0.04	4.61 ±0.04	5.59 ±0.06	6.58 ±0.08	7.92 ±0.11	1.02 ±0.02	0.99 ±0.02	0.98 ±0.02	0.99 ±0.02	1.34 ±0.03	0.29 ±0.01

The split times showed a stronger correlation to race result for men rather than for women, except for run-in, that was significantly correlated to race time for women only (Table 2).

Table 2. Correlation characteristics of the final results and the interval times.

	H1/H2	H2/H3	H3/H4	H4/H5	Run-in	MaxInterV	MeanClearT
Men (n=8)	.32	.90**	.97**	.77*	.22	-.74*	.73*
Women (n=8)	.69	.80*	.77*	.75*	.85**	-.83*	-.25

(*P<0.05, **P<0.01)

Women showed an expected trend of quicker hurdle clearance than men (Mean: 0.29±0.01 s vs. 0.33±0.02 s, respectively). Interestingly, the mean hurdle clearance time (MeanClearT) showed a significant correlation to final result for men (r = .73, P<.05) but not for women (r = -.25).

DISCUSSION: The present data shows that the race pattern for indoor 60-m hurdles is similar to the beginning of the outdoor 100-m/110-m hurdles. Women tended to reach their maximal interval speed later than men, which is in line with the previous results from outdoor competitions (Brüggemann & Glad, 1990; Brüggemann et al., 1999). The observed significant relationship between the mean hurdle clearance time and race result for men only suggests that hurdle clearance plays a more important role in race performance for men than for women. This is likely due to a relatively lower hurdle height in women's sprint hurdles enabling faster clearance and smaller loss of horizontal velocity (Salo et al., 1997). The longer run-in distance in women's 60-m hurdles, as compared to men (13m vs. 9.72m, respectively), may further favour the role of high 'pure' running speed as an indicator of race performance in women's indoor sprint hurdles. This was supported by the significant correlation between the run-in time and race result for women.

CONCLUSION: The racing patterns of world-class hurdle sprinters in 60-m hurdles were found to be similar to the beginning of the race of those reported in 100-m/110-m hurdle competitions. Hurdle clearance seems to play more important role in competition performance for men than in women at the world-class level.

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EFFECTS OF AN ANGLED STARTING BLOCK ON SPRINT START KINEMATICS

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KEYWORDS: Sprint start kinematics, angled starting block prototype

INTRODUCTION: The sprint start mechanics have changed over the years with the invention of the starting blocks. The most recent modification of the starting block developed by Gill Athletics was to create an outward angle of the foot position in the block in order to match the .28 rad. (16 deg.) oblique axis of the ankle. This study examined the effects of an outward angled foot position in a starting block prototype on the first four steps during sprinting kinematics.

METHODS: Six female sprinters in the middle of their outdoor season volunteered from the Indiana State University Track team read and signed an informed consent before they performed sprint starts from a standard block and a Gill angled starting block prototype. The subjects' mean age, height, and weight were 20.0 ± 0.9 yrs, 1.65 ± .08 m, and 59.9 ± 5.0 kg. The prototype starting blocks were made with an outward, .28 rad. angle of the foot pad in order to permit full foot force to be applied perpendicularly to the block pad. From the height, the blocks were adjusted such that the lead knee was placed at a 1.75 rad angle and the subjects were permitted to practice using the new blocks for a one hour period over one week. Prior to testing, subjects jogged 400 m, and then performed their regular sprinter stretches. Each subject performed 3 maximal sprint starts for 30 meters using either the standard block or the angled block followed by the other block shown in Figure 1.

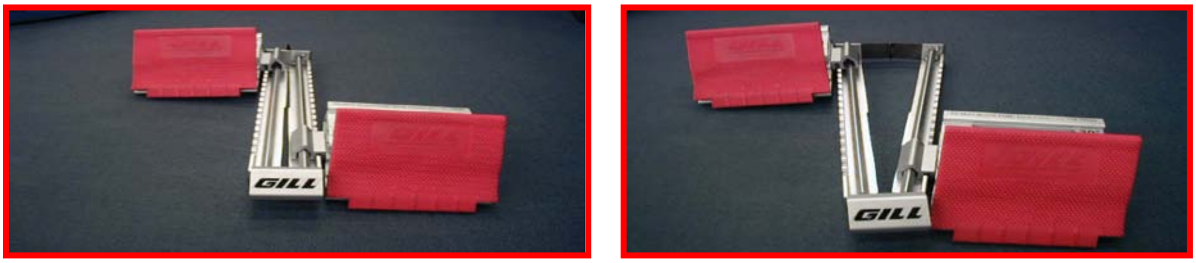

Figure 1. Traditional starting block & Gill angled starting block prototype

All sprint starts were recorded at 60 Hz with a .001 s shutter from side, right rear, and front views, 16 body markers were identified, and a calibration cube was placed in the middle of the 4 step field of view. Markers were digitized, transformed using the 3D direct linear transformation (DLT), and digitally filtered at 10 Hz using the Ariel APAS 2008. Kinematic variables of step length, total body CM horizontal velocity, and the block clearance times were calculated and analyzed with an ANOVA with repeated measures.

RESULTS AND DISCUSSION: The analysis found significant differences ($p<.001$) between the first four running step lengths but not between the type of block, which is shown in Figure 2. Also, the body's CM velocities were significantly different for the step factor where the 4th step was fastest, which is illustrated in Figure 3. Research conducted by Guissard (1992) found that by adjusting the block's medial/lateral angle to replicate the oblique ankle angle on the starting block significantly resulted in sprint start velocities that were .03 m/s faster in the first 4 steps. Their study found the angled blocks elicited a slightly higher non-significant average linear velocity of 4.63 m/s as compared to 4.60 m/s for the standard block. In the present study, the block clearance time was measured when the back foot passed the front part of the block. Non-significant ($p=.334$) mean block clearance times were .482 ± .050 s for the angled block and .470 ± .038 s for the standard starting block, shown in Figure 4. It was reported by Ozolin (1988) that sprinters take 0.3 to 0.4 seconds to react to the starting gun and clear the blocks. The longer time for force being applied against the angled block provided could produce a larger impulse even with similar forces, and may be responsible for the slightly higher CM velocities seen in the accelerative phase (2nd, 3rd, and 4thsteps).

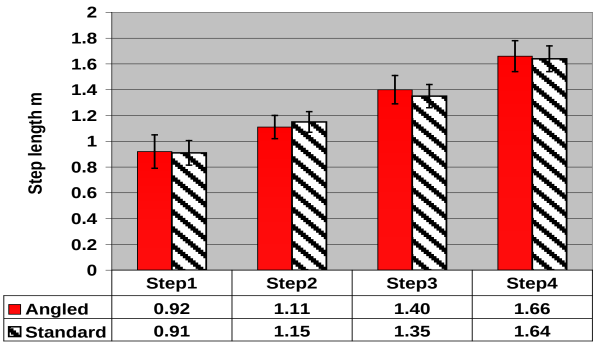

Figure 2: Step lengths using blocks

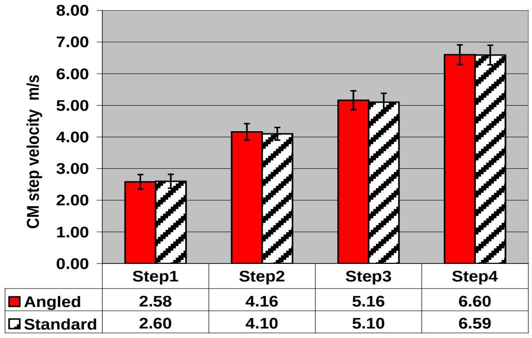

Figure 3: CM step velocity using blocks

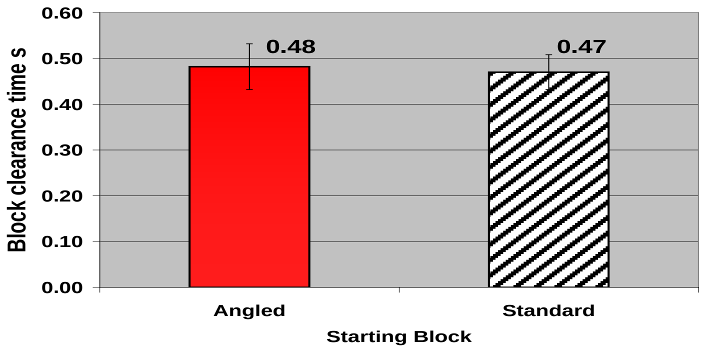

Figure 4: Block clearance time

CONCLUSIONS: A .28 radian oblique angle on the block allowed the athletes to apply their foot force perpendicularly against the block for a longer time period. An increase in the runners' explosive impulse coming out of the blocks produced longer block clearance times, and slightly higher maximal velocities during the accelerative phase (steps 2, 3, and 4). Although the angled block did not produce greater velocities during the first step, it did facilitate an effective transition to form running with slightly higher velocities which were non-significantly different in the latter steps measured. According to research conducted by Stevenson (1997), a longer stride resulted in a higher linear velocity at take-off, but hindered block clearance time. The present study found that the angled starting block produced slightly longer step lengths for 3rd and 4th steps, which resulted in higher linear velocities but slower block clearance times. Overall, the angled block was equally effective as the standard block with only slight, non-significant improvements in the runners' velocity. Our study's relatively slow .017 s frame duration may have not provided a fast enough sampling rate, to measure minor adjustments in timing and velocities which can occur in time intervals less than .001 s. Although, these slight differences may appear to be insignificant, they may be beneficial for coaches and athletes, in sprinting events where the margin of winning or losing is measured in milliseconds.

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COMPARISON OF INSIDE CONTACT PHASE AND OUTSIDE CONTACT PHASE IN CURVED SPRINTING.

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KEYWORDS: Curved sprinting, centripetal force, body lean

INTRODUCTION: One of differences of running between in straight path and in bent path is the body lean inward. When athletes run in the bent path, athletes are influenced by centrifugal force. Athletes produce the medio-lateral component of ground reaction force (GRF) with inclining the body inward, to balance the centrifugal force. The centripetal force can be estimated by

$$F=mv^2/r \tag{1}$$

where F is the centripetal force, m is the mass of the body, v is the running velocity, and r is the radius of track. We can see from this equation, the influence of the running velocity on centripetal force is strong. So we can't ignore the centripetal force in sprint race (i.e. 200m dash). The body lean angle increase, as the running velocity increase. Athletes must change the running direction along the bent path. One step of running consist of the flight phase and the contact phase. It is impossible to change running direction during the flight phase. Therefore it is expected to change running direction during the contact phase. It was said the functions of inside and outside legs were asymmetrical (Stoner and Ben-Sira, 1979, Hamill et al, 1987). But there are no study to compare the inside foot contact phase with the outside foot contact phase with in relation to the centripetal force and the body lean angle. The purpose of this study is to compare the inside (left) foot contact phase with the outside (right) foot contact phase with in relation to the centripetal force and the body lean angle. Then we would obtain new knowlege about curved sprinting.

METHOD: Six male sprinters (age: 20.7±1.5years, height: 171.9±5.3cm, body mass: 63.7 ±4.3kg) participated in this study. Each subject ran about 60m with maximal effort on the bent path 3 times (4th lane, radius: 41.85m) of an official outdoor track with his own spike shoes and with 41 reflective makers. The running motion was recorded using 3D motion capture system (Vicon) with 10 infrared cameras (125Hz). The experiment was conducted after dark because of the data collection with the infrared motion capture system. The obtained 3D coordinate data were smoothed using butterworth type digital filter, with the optimal cutoff frequencies (Winter, 1990). The variables, 1) average horizontal velocity of center of mass runner's body, 2) contact time, 3) average centripetal force, 4) body lean angle at foot contact and toe off, were compared between two contact phases of inside and outside using a paired t-test. The centripetal force was calculated using formula (1). The body lean angle was defined as the angle between the vertical line and the line from the center of mass of foot to that of body in frontal plane. Hip adduction-abduction angles were also obtained and compared between two phases.

RESULTS: There were no significant differences in the average horizontal velocity, and the average centripetal force between two contact phases of inside and outside feet. The outside foot contact time was shorter than the inside one (right: 0.112s, left: 0.125s, $p<.01$). The body lean angles had significantly different at toe off (right: 13.6deg, left: 2.05deg, $p<.01$). During inside foot contact phase, the body lean angle was decreasing, while during outside, the angle was increasing (Figure 1).

DISCUSSION: The previous study (Stoner and Ben-Sira, 1979) showed the average velocity of inside foot contact phase was slower than the outside. But our results didn't show the difference between two phases. It because there was no remarkable difference in the velocity, the centripetal force had no difference. The inside foot contact time was longer than the outside foot contact time, and it agree with result of previous study (Stoner and Ben-Sira, 1979). It had been predicted that to accomplish curved running, athletes may run along bent path for adjusting running direction by adducting inside leg and abducting outside leg. But the inside hip abducted (Figure 1). Our study's radius was 41.85m, it was sufficient radius not to adjust running direction by adduction. As expected, the body lean angle was increasing with hip abduction during outside foot contact phase (Figure 1). It may be accomplished curved running by outside leg abduction. Hamill, et al (1987) described "modifications in body position, and thus the lower extremity, are, strictly speaking, a function of the runner's velocity." We think the centripetal force is also functions of the runner's velocity. But in this study, there were no difference in the centripetal force. It is required more advanced research including measurement of GRF to explain the curved running mechanism.

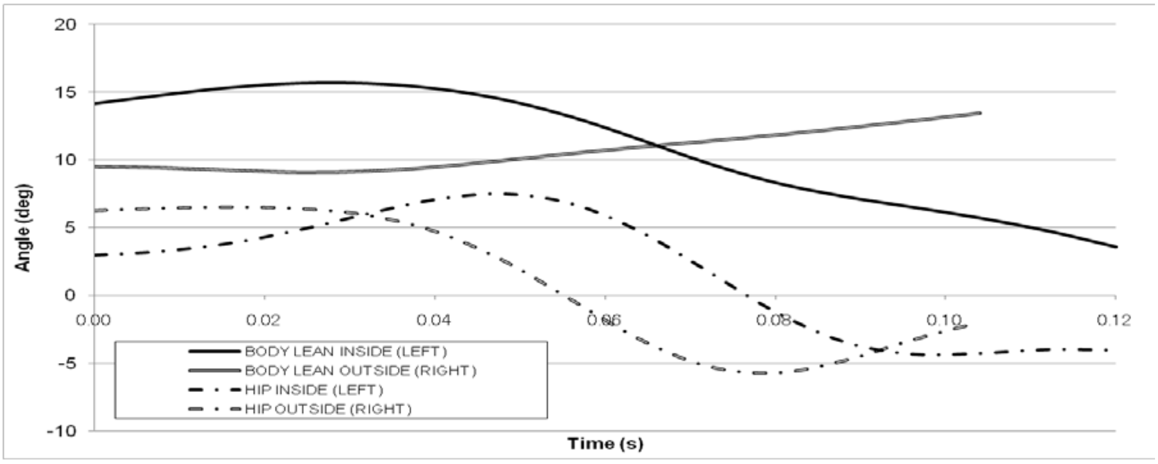

Figure 1. The typical example of the change of the body leaning angle and hip adduction-abduction angle during foot contact phases. In the body leaning angle a positive value indicates the body incline inward, and a negative value indicates the body incline outward. In the hip add-abd angle, a positive value shows adduction, a negative value shows abduction.

CONCLUSION: This study indicated the differences between the inside foot contact phase and the outside in curved sprinting. The body lean angle was decreasing during inside foot contact phase and increasing during outside foot contact phase. It seems that more advanced study is needed to explain the curved running.

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TREKKING POLE FORCES DURING DOWNHILL WALKING

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This study examined gender differences when hiking downhill with trekking poles. Fourteen men and thirteen women were recruited who had hiking and poling experience. Integrated pole forces were examined over two pole strikes (left pole followed by right pole) prior to and during a stance phase of a step. Total pole force was compared between gender, as well the percent of pole force during the actual stance phase of the step. Left and right pole strikes were also examined for symmetry. Men generated a greater combined pole force than women (0.61N/kg vs. 0.48N/kg) but the differences were not statistically significant. During the stance phase, 48% of the combined pole force occurred for men, but only 35% of the pole force was noted for the women. Pole forces were less symmetrical for the women as well, although also not statistically different. Similar total pole forces between gender with less pole force during stance phase indicates pole walking technique differences rather than a lack of upper body strength, for women, who previously demonstrated less footfall force changes when walking with poles than without, in comparison to men.

KEYWORDS: gait, hiking, gender

INTRODUCTION: The effects of trekking pole use while walking downhill have been well established including significant decreases in foot plant forces (ground reaction forces (GRF), braking forces (BF)) and muscle activity of the lower extremity (Bohne & Abendroth, 2007; Schwameder, Roithner, Muller, Niessen, & Raschner, 1999, Willson, Torry, Decker, Kernozek & Steadman, 2000); however, the mechanism behind these effects is not well understood. Theoretically, with pole use, a portion of the force acting on the body is transferred to the upper extremity, thus decreasing the forces acting on the lower extremity (pole loading). Previously, it has been demonstrated the effects seen with hiking pole use are the result of pole loading during the stance phases of walking. This research also indicated that men appear to use poles more effectively than women when walking downhill (Abendroth, Dixon, & Bohne, 2009). However, it remains unclear whether the lesser force reduction noted in the previous study was due to women being unable to load the poles as effectively as men. It has been suggested that women lack the necessary upper body strength to transfer forces to the upper extremity with pole use.

Therefore, the purpose of the current study was to compare poling forces between men and women during downhill walking, before and during the stance phase of a step. Specific goals were to examine pole loading during a full stride (left and right pole strike), beyond a specific stance phase, between gender. It was hypothesized that significantly different pole load forces, per body kilogram, would be seen between men and women. This would correlate with previously examined decreases in foot plant kinetics, thus supporting the idea that women lack in upper body strength, and so unweight to a lesser degree than men are able to. If women do in fact load the poles in a similar manner as men, then pole walking techniques may be the reason behind the differences in subsequent foot force differences between gender. A secondary goal of the research was to examine pole force symmetry, when walking with two poles. Pole force asymmetry may also indicate walking technique differences.

METHOD: Twenty-seven healthy volunteers with previous hiking and pole use experience were recruited, and all signed informed consents (14 men, 13 women: age 39 ±12; 43±13 and mass 82.2 ±6.5; 61.2±7.1 kg, respectively). Approval for the study was obtained from the University IRB.

Participants were instructed to wear their preferred hiking shoes, and then assigned to walk in a predetermined, counterbalanced order which included a no pole (NP) and a pole (P) condition. Participants walked at a self-selected pace, although walking speed was held constant between conditions using a Brower timing system. Additionally, the participants used a self-selected poling technique. Pole length was self-selected due to the experience of the participants. Participants completed 10 successful trials (complete force plate contact during "natural stride") in each condition.
A wooden ramp with ascending and descending 20 degree slopes (3.3 m descent) was used to simulate a hiking experience. A Bertec force plate (40 cm x 60 cm; 1000Hz) mounted flush in the down slope portion of the ramp was used to collect ground reaction and braking forces (GRF/BF). Bertec instrumented Leki trekking poles (1000 hz) were used to measure pole load. Pole forces were collected via Bertec Acquire software. All data were collected simultaneously over 3 second intervals during the downhill portion of the walk. Pole forces were examined over two pole strikes, consisting of a left and right side strike, prior to and during the foot strike on the force plate. The pole forces were integrated and averaged over ten trials, for both sexes. The loads were then normalized to body mass (N/kg) for each participant. Data collected prior to, during and after the stance phase (when the participant's foot was in contact with the force plate), were used to perform analyses between pole forces and gender.
Two pole loads, consisting of the left and right strike prior to and during the collected left side stance phase of the foot strike were synchronized with the stance phase of the force data. The average loads generated for both poles during this stance phase were integrated and averaged over the ten trials. Left and right integrated pole forces were compared with each other, and between gender. Percent of poles forces used during the stance phase of one step divided the integrated pole force was also calculated and compared between gender. Statistical analyses were performed using SPSS(version 16.0). Comparisons were made between men and women for the pole forces using a mixed 2-way ANOVA (pole side x gender). Percent of pole forces during the stance phase were compared using a Chi-Square test. Statistical significance was set at an alpha = .05 for comparisons. Statistical power was calculated to be 80%, based on the selected sample size and an effect size of 0.7.

RESULTS: Figure one demonstrates the average (SD) for left and right pole use (with a left foot strike on the force plate), between gender, as well as the amount of force generated by pole use only during the stance phase of the foot strike. The men, on averaged produced integrated pole forces of 0.30 and 0.31±15 N/kg for left and right pole strikes. Women produced 0.22 and 0.26±.14N/kg for their respective pole strikes. For men, the stance phase pole forces were 48.5% (0.29 N/kg) of the total pole force used when the left and right pole forces were combined. The women used 35.2% (0.17N/kg) of the combined pole force during their stance phase.
The 2 way mixed ANOVA demonstrated no significant main effects for gender (F= 1.49, df = 1, p= .234). The main effect of pole symmetry was also non-significant (F=2.95, df = 1, p = .098). Interaction between pole symmetry and gender was not statistically significant as well (F= 0.67, df=1, p=.42). Effect sizes (ES) were examined between gender for practical significance. A moderate effect size was noted for the combined integrated pole force (ES = 0.44) between gender. Pole symmetry ES for women was moderate at .36 but men's were low at .10. A significant difference between the percent of pole force use during stance phase with men and women (p=.035) was noted, using a Chi Square.

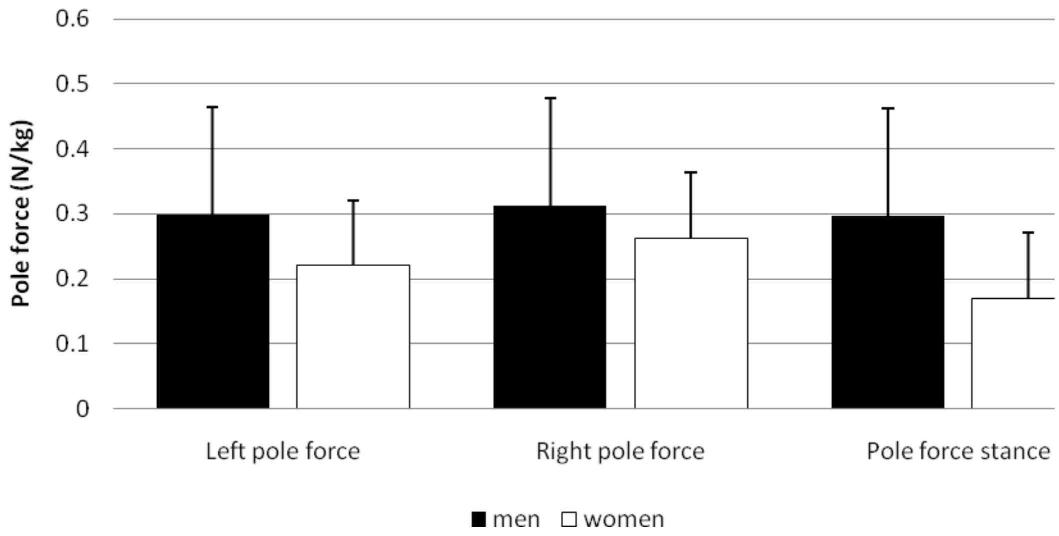

Figure 1. Forces applied to left and right poles during a pole plants (beyond and during stance phase) as well as total pole force used during stance by gender.

High variability within gender was noted for poling forces. Patterns were examined for age or experience with trekking poles, in relation to the pole forces, but no patterns could be discerned.

DISCUSSION: The primary purpose of this study was to examine gender differences with trekking pole loads prior to and during a stance phase of a downhill hike. This would better discern the role of using poles for unweighting vs. stability pole walking technique differences. Men do load the poles greater than women, even when normalized. However, this difference, when examining two full pole strikes, is relatively small. For a typical 70 kg male, the poles would be loaded with an average force of 42.7 N between the two poles (approx. 6% BW). The typical 50kg woman would load her poles with an average force of 24N (approx 5% BW). However, for men, 48% of the pole force was administered during the stance phase; while only 35 % of the pole force is noted during the stance phase for women. This may support the idea that women are not using to poles predominately to unweight but for other reasons, intentional or not. Previously it has been hypothesized that women employ an altered technique due to a lack of upper body strength to effectively unweight themselves during pole plants. However, the current results may indicate that there is a difference in technique of women while walking with poles, but may not be related to an upper body strength issue but rather to increase stability during the descent.
Previously reported foot strike data using the force plate included women, on average, demonstrating similar forces when using poles in comparison to no poles, except for peakFz. Men demonstrated decreased peak forces in both vertical ground reaction force (VGRF) and BF, but the integrated forces were similar during the poling condition and non-poling conditions. Specifically, an average reduction of peak VGRF of 1.1 N/kg for men, and only 0.09 N/kg for women was noted. The average braking forces were reduced for men by 0.46 N/kg while women increased their braking forces with poles, by 0.12 N/kg. (Abendroth, Dixon, & Bohne, 2009)
Pole load symmetry was noted as a secondary factor; women demonstrated less symmetry with left and right pole strikes than did men. It is possible, that while this asymmetry was small, it may be a further indication of technique differences between gender when walking with trekking poles. Additionally, dominant were not controlled and may lead to an explanation of the asymmetry. There is also a possibility that the velocity between walkers may play a role (although held constant within each participant), however, was not controlled in the current study.

Lastly, high variability among gender limits the strength of the conclusions. It was also noted that age and experience with trekking poles did not correspond to the pole loads noted.

CONCLUSION: Gender alone does not appear to a predominant indicator in effective pole use. However, technique differences may be more responsible for lesser transfer of forces from the lower extremities by women, than upper body strength, since women seem to load the poles at similar levels as men, when normalized to body mass. Examining the location and timing of pole plants, relative to foot strikes, may further indicate techniques differences with trekking pole walking, between gender. However, upper body strength was not measured in the current study and its role can be further analyzed.

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LONGITUDINAL KINEMATIC CHANGES WITH THE DIAGONAL STRIDE IN HIGH-SCHOOL GIRL CROSS-COUNTRY SKIERS

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The purpose of this study was to describe longitudinal kinematic changes associated with the diagonal stride in high-school girl skiers. An emphasis was placed on the poling phase of the movement and the stretch-shortening movements that can observed at the elite level. Four high-school athletes were videotaped once each year for 4 years. Angular relationships and kinematics variables were evaluated. Most athletes showed steadier trunk angles, and more consistent elbow extension in later years. Elbow angles decrease at the beginning of the poling phase, stretching the elbow extensors in the initial part of the propulsive phase. All athletes progressively increase knee angle during the glide phase but do not show a short but significant flexion just prior to pole plant and initiation of the "kick", which was exhibited by the model. Coaches can benefit from angular analyses as their emphasis in training is often related to body positions during various phases of the movement.

KEY WORDS: cross-country skiing, kinematics, performance, technique.

INTRODUCTION: Cross-country skiing has undergone a series of changes in the past thirty years, from the introduction of the skating technique to the new sprint competitions that have be seen since the 2002 Olympic games. However, the traditional diagonal stride (classical technique) is still very popular and usually taught first to the novice skier.

Biomechanical markers of performance (such as stride length, stride rate, center of mass displacement and velocity) in elite racers have been identified for all cross-country skiing techniques. Studies have shown stride length to be a better determinant of performance (average speed over a race) for any given technique in comparison to stride rate (Bilodeau et al., 1996; Smith et al., 1989; Smith & Heagy, 1994).

The diagonal stride can be described as the arms and legs moving in a diagonal fashion, with a push-off performed by the arm and leg of the contralateral side (Nilsson et al., 2004). The focus of diagonal stride investigations evaluated temporal characteristics (such as time of glide and recovery) compared to the propulsive phase at the elite level (Bilodeau et al., 1992; Dufek & Bates, 1987; Komi, Norman, & Caldwell, 1982; Lindinger et al., 2009). However, technical changes associated with the developing high-school athlete have yet to be addressed. The purpose of this study was to describe selected longitudinal kinematic changes associated with the diagonal stride in high-school girl skiers. An emphasis was placed on angular changes associated with the poling phase of the movement.

METHOD: Over a 4-year period, more than 30 female high-school skiers from the Duluth, MN area participated in this project, but only 4 skiers were involved in each of the data collecting sessions and are described here (age = 13.1 ± 1 years in Year 1). The participants were also compared to a "model" subject (age = 17), a high-school girl skier selected by various coaches to have an excellent technique. All subjects signed a consent and a parental permission was obtained in accordance with the University of Minnesota IRB procedures.

Participants proceeded through a self-guided warm-up and performed the diagonal stride along a straight and flat track (0% grade) of approximately 150 meters in length groomed specifically for this project. The subjects used their own equipment, which changed in size from Year 1 to Year 4, and the coaches waxed the participants' skis prior to data collection. Temperatures during data collection were remarkbly similar each year (-6 to -8 °C), but snow conditions in Year 4 were described as "icy", necessitating a Klister "kick" wax for the participants. The participants were asked to ski at race pace, trying to maintain a constant speed through a 20-meter marked zone of the track. Two GEN-LOCK synchronized 60-Hz cameras (JVC-TKC1380) were used to videotape the performances. These cameras were

placed on specialized tripod heads, allowed panning and tilting since movement occurred over a large calibration area (ViconPeak, Centennial, CO).
One cycle for each performance was randomly chosen for digitizing. The video images were manually digitized using a 23-point, 16-segment model (DeLeva, 1996), and the position-time data from each camera processed with Peak Motus 8 using the Linear Transformation Technique (DLT) to create the 3-D model (Abdel-Aziz, & Karara, 1971). Cycle length (CL), velocity (v), and percent poling phase time (%PP) were calculated. Angular relationships related to trunk, pole, elbow and knee were also calculated.

RESULTS: Unfortunately, no data were collected in Year 3 due to poor snow conditions. The Model skier (M) is described for Year 2 only (the last year she skied for this project). There is no particular order to the listing of each individual subject. Table 1 describes performance markers for each athlete. Most skiers had a poling phase (PP) relative time between 30 and 37%, with S1 and S2 showing the largest variation between years.

Table 1. Selected kinematic markers of performance

	Cycle Length (m)					Velocity ($m.s^{-1}$)					% Poling phase Time				
	M	S1	S2	S3	S4	M	S1	S2	S3	S4	M	S1	S2	S3	S4
Year 1		5.88	5.54	4.03	4.85		3.96	3.70	3.41	3.31		30.3	32.2	36.6	36.4
Year 2	6.01	6.31	5.93	5.09	5.57	4.74	4.12	4.68	4.02	3.88	31.6	28.3	31.6	35.5	32.6
Year 4		5.08	5.08	4.63	4.42		3.71	4.12	3.56	3.31		35.4	29.7	35.0	35.0

Trunk and elbow angles changes during the poling phase are important factors associated with the overall performance of the skiers. All skiers showed a steadier trunk angle in the later years (slope is more vertical), with less angle variation through the plant. However, all skiers (except M and S2 in Year4) seem to bring the trunk up before the end of PP as can be observed in Figure 1.

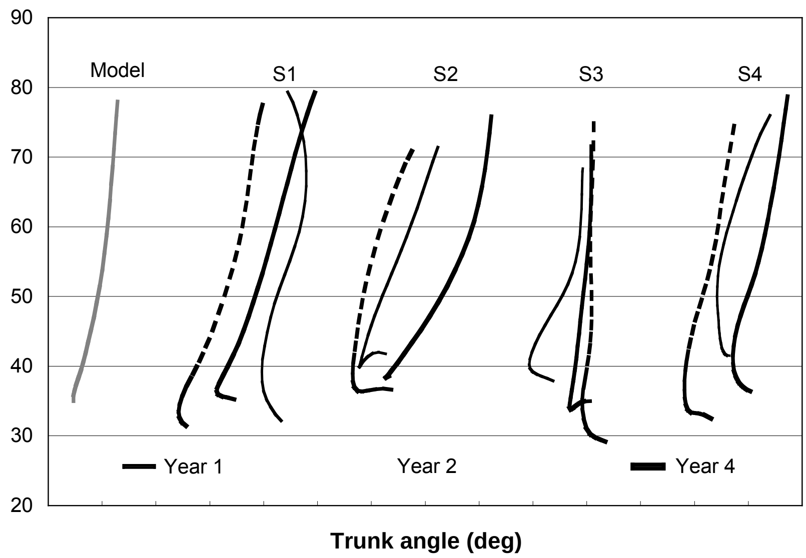

Figure 1. Individual trunk angle changes during the poling phase of the diagonal stride.

The elbow angular changes exhibit a “stretch-shortening” movement by which the athlete initially flexes the elbow, stretching the triceps, and subsequently extends through the poling phase. This can be observed in all skiers in one of the various years (Figure 2), but it is not consistent through the years. Only S1 and S2 showed a progression from Year 1 to Year 4.
In the glide prior to pole plant on the contra-lateral side, the knee angle is held at a relatively constant angle as can be observed in M between 25 and 50% CT in Figure 3. The poling

phase occur at approximately 50% of cycle time (CT). Only M truly exhibit a slight extension prior to pole plant, which also initiates the kicking action (pushing down on the ski).

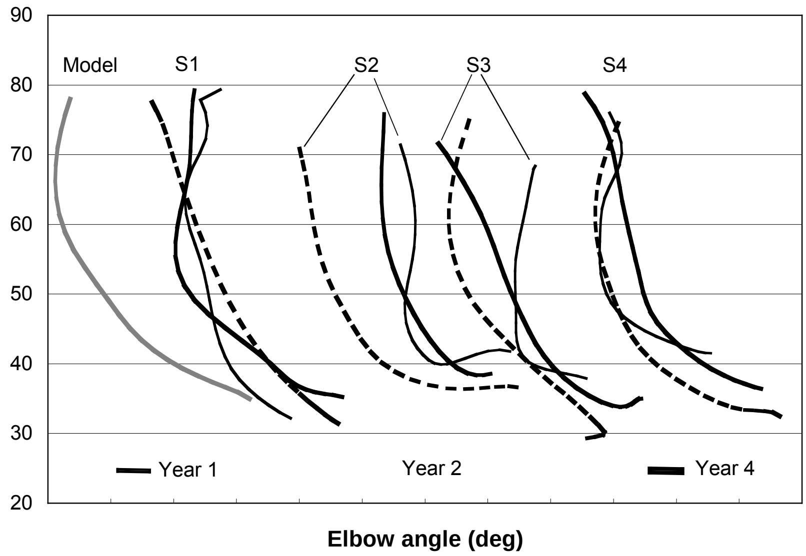

Figure 2. Individual angular changes during the poling phase of the diagonal stride.

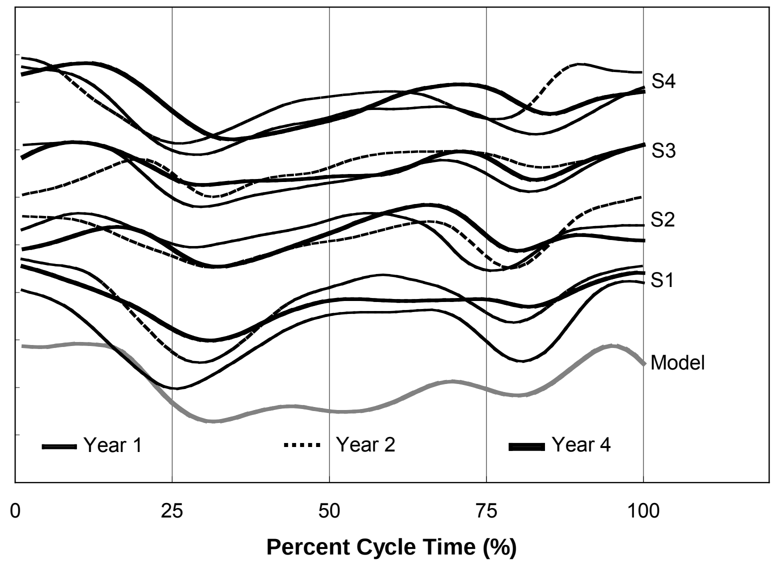

Figure 3. Knee angular changes for one cycle

DISCUSSION: The athletes showed progress from Year 1 to Year 2, but conditions in Year 4 may have contributed to the decline in performance-related markers. Observed PP relative time was similar to elite levels (Bilodeau et al., 1992; Dufek & Bates, 1987). However, angle relationships during the poling phase expressed marked changes. Athletes showed more consistent variations in their trunk angle, expressed by the the more vertical curves of Figure 1, with a trunk angle variation of approximately 10 degrees in Year 4 from 15 degrees in Year 1. All skiers (except M and S2 in Year 4) bring their trunk up before the end of the poling phase. While this action is very short in absolute time, it suggests that the athlete may attempt to use trunk extension to extend the poling phase. It is not certain whether this action can be beneficial to maintaining speed and whether it positively affects performance. Elbow angles variations suggest that the athletes initially flex the elbow at the beginning of the poling phase, stretching the extensor muscles before engaging into forceful elbow extension (Figure 2). This small flexion was reported by Lindinger et al. (2009) during uphill roller skiing with elite skiers, but this movement peculiarity is not commonly observed at the high-school level.

Knee angle should remain fairly constant through the glide phase (from about 25 to 50% of CT). Only M exhibits the slight knee extension followed by the small and quick knee flexion, which allows a vertical force to compress the ski against the snow, initiating the “kick” (Smith, 2000). This angular change was not observed in the other skiers. Instead, there is an extension of the knee throughout the glide phase, suggesting a constant movement toward the initiation of the “kick”. Only S3 in Year 4 keeps the knee angle constant through the glide phase. This constant increase in knee angle raises the center of mass, but the lack of a sudden knee flexion at the initiation of the kick suggests these athletes must find another strategy to accelerate the center of mass down to compress the ski against the snow.

CONCLUSION: This project described longitudinal changes associated with the diagonal stride in cross-country skiing. In particular, it focused on the non-elite athlete at the high-school level. Most athletes showed improvement from Year 1 to Year 2, but not necessarily from Year 2 to Year 4. Measurable changes can be observed in individual athletes, and technique deficiencies can be detected from year to year as evidenced by angular changes. Environmental differences will affect performance markers such as speed, stride length and rate, but body positioning measures such as angular relationships tend to be independent of these environmental changes and could be helpful to the development of the young skier.

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TEMPORAL METHODS TO ESTIMATE THE DISPLACEMENT OF A CURLING ROCK: COMPARISON BETWEEN COMPETITIVE AND RECREATIONAL CURLERS

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The purpose of this study was to examine different methods used in curling to estimate the total rock displacement. A group of competitive (n=8) and recreational (n=8) curlers each delivered a total of 16 rocks, both guards and draws. Interval times for each delivery were measured from the back line to the near hogline and from the near hogline to the far hogline, and the average speed after release and the total rock displacement were determined. Pearson product moment correlations were calculated among the variables for each participant. The results of the study indicated that the various timing methods to estimate the total displacement of the curling rock are appropriate for competitive curlers, but may not provide accurate estimates for all recreational curlers.

KEYWORDS: curling, delivery, competitive, recreational

INTRODUCTION: One of the fundamental skills in the sport of curling is the delivery, in which the objective is to "throw" the rock so that it will eventually come to rest in a specific position at the opposite end of the sheet, or so that it will contact and remove an opponent's rock from play. Curlers attempt to deliver the rock with the appropriate speed and direction. The rock must be released with the required for the type of shot being played, whether it is a guard, draw, or take-out. Also, the rock must be delivered in the appropriate direction to allow for the proper amount of curl to complete the required shot (Bradley, 2009; Buckingham, Marmo & Blackford, 2006). It is the curling delivery which has the correct combination of speed and direction which will be most successful.

To assist curlers in successfully playing the required shot, various timing methods are often used to estimate the speed the rock is traveling and where it may eventually stop. This involves measuring the time required for the rock to travel specific distances as it moves down the ice as indicated by fixed lines on the playing surface. Timing will help determine whether or not it is necessary to sweep in order to extend and/or manipulate the path of the rock (Behm, 2007). One of these measurements is the time interval from the back line to the near hogline. A second frequently used measurement is the time interval between the near hogline and the far hogline (Figure 1.). Curlers measure these times during competitive play using a stopwatch.

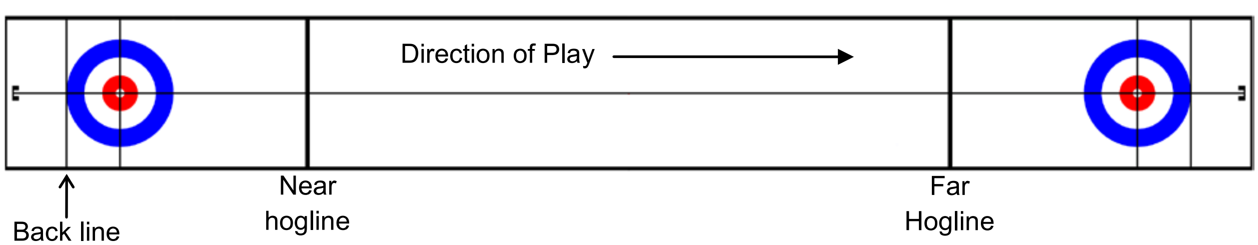

Figure 1. Time interval measurements in curling.

Curlers of all ability levels use these timing methods to estimate the "speed" of the ice and where the curling rock will eventually stop, however, no studies have examined these timing methods to determine their effectiveness or accuracy. In addition, it is currently not known whether or not these timing methods are appropriate for use by curlers of all levels, as the quality of the delivery often seen in less skilled curlers may influence the effectiveness of the timing. Therefore, the purpose of this study was to examine the different timing methods used in curling to estimate the total rock displacement. Comparisons were also be made between competitive and recreational level curlers.

METHOD: Sixteen curlers were recruited, eight competitive and eight recreational. The competitive curlers (mean age = 27.0 yrs; mean years of curling experience = 17.3 yrs) were players in the Major League of Curling in Thunder Bay, and played or practiced more than two times per week. This group included a number of athletes who have competed at provincial, national, and international events. The recreational curlers (mean age = 33.3 yrs; mean years of curling experience = 2.9 yrs) were individuals who played for various club teams in the city, and played or practiced one time per week or less. Ethical approval was received from the Lakehead University Research Ethics Board prior to data collection.

Participants were required to deliver sixteen rocks, eight draws and eight guards. Draw shots were those that stopped in the house; guard shots were those that stopped in front of the house. The deliveries were completed in random order, with equal numbers of in-turn and out-turn rotations. Timing of each curling rock was completed using a Brower wireless timing system (Draper, Utah). Interval times for each delivery were measured from the back line to the near hogline, and from the near hogline to the far hogline. In addition, a beam was located one meter past the near hogline and was used to determine the average speed of the rock after release. The total rock displacement was measured from the near hogline to the final stopped position using a measuring tape.

Data collection took place at the same curling venue but on multiple days and on different ice sheets. Because of this, it was not possible to control for the speed of the ice and ensure consistent test conditions for all participants. This prevented the data among the competitive and recreational curlers from being grouped for analysis. Pearson product moment correlations were calculated among the variables for each individual participant.

RESULTS AND DISCUSSION: Temporal data indicating the range (minimum and maximum) of the measured times and the speeds for rocks that were successfully delivered as guards or draws for all participants are presented in Table 1. Guards, which travel a shorter distance because they come to rest in front of the house, have longer back line to hogline times, slower rock speeds at release, and longer near hogline to far hogline times. In comparison, draws travel a further distance because they stop in the house. Accordingly, shorter back line to hogline times, faster rock speeds at release, and shorter near hogline to far hogline times are observed.

Table 1. Temporal and Speed Data

	Guard		Draw	
Variable	Min	Max	Min	Max
$\text{Time}_{(\text{back line - near hogline})}$ (sec)	3.44	4.11	3.38	3.83
Average speed at release (m/s)	1.89	2.22	2.00	2.33
$\text{Time}_{(\text{near hogline - far hogline})}$ (sec)	14.00	18.49	11.42	14.35

The results of the correlational analysis are presented in Table 2, outlining the range (minimum and maximum) of r values among the participants in each group. Statistically significant correlations were seen for all the competitive curlers among the temporal and displacement variables. Shorter back line to near hogline times were significantly correlated with faster average speeds at release, shorter near hogline to far hogline times, and greater rock displacements. Both the average speed at release and near hogline to far hogline time were also strongly correlated with rock displacement. For the recreational curlers, the results of the correlational analysis were more variable. While some of the recreational curlers demonstrated significant correlations for their deliveries similar to the competitive curlers, other recreational curlers did not. In particular, non-significant r values were seen for the three correlations associated with the variable average speed at release.

Table 2. Correlational Analysis

Correlation	Competitive Min	Competitive Max	Recreational Min	Recreational Max
$\text{Time}_{(\text{back line - near hogline})}$ - Average speed at release	-0.85 [a]	-0.95 [a]	-0.46	-0.93 [a]
$\text{Time}_{(\text{back line - near hogline})}$ - $\text{Time}_{(\text{near hogline - far hogline})}$	0.90 [a]	0.95 [a]	0.70 [a]	0.93 [a]
Average speed at release - $\text{Time}_{(\text{near hogline - far hogline})}$	-0.85 [a]	-0.96 [a]	-0.41	-0.97 [a]
$\text{Time}_{(\text{back line - near hogline})}$ - Rock displacement	-0.87 [a]	-0.94 [a]	-0.66 [a]	-0.94 [a]
Average speed at release - Rock displacement	0.89 [a]	0.98 [a]	0.23	0.96 [a]
$\text{Time}_{(\text{near hogline - far hogline})}$ - Rock displacement	-0.93 [a]	-0.99 [a]	-0.89 [a]	-0.98 [a]

[a] $p < 0.05$

A reliable delivery should be the objective of all curlers, whether they participate at a competitive or recreational level. Having the ability to deliver the rock so it is traveling in the intended direction with the appropriate speed will increase the chances of successfully completing the required shot. The results of this analysis indicate that statistically significant relationships were seen among the temporal variables associated with rock displacement in competitive curlers. These curlers demonstrate a high degree of reliability in their delivery and are capable of making the small adjustments that are necessary for the variety of shots being played. For recreational curlers, however, the results were less consistent. Some recreational curlers have the ability to deliver the rock reliably, but others do not. In particular, it was the three correlations involving the average speed at release which resulted in the non-significant r values among the recreational curlers. This suggests that curlers of this ability level may have difficulty delivering and releasing the rock consistently with the correct speed and adjusting to various shots. It is important to note, however, that the correlations between the time from the near hogline to the far hogline and the rock displacement were statistically significant for all participants in both groups. This time interval may be the most effective for estimating the total rock displacement for all curlers.

CONCLUSION: The results of this study indicate that the various timing methods to estimate the total displacement of the curling rock are appropriate for competitive curlers, but may not provide accurate estimates for all recreational curlers. It is important for curlers of all levels to develop a reliable delivery in order to be successful.

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Poster Session 4

THE EFFECT OF DEPTH ON THE DRAG FORCE DURING UNDERWATER GLIDING: A CFD APPROACH

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KEYWORDS: swimming, gliding, CFD, drag force.

INTRODUCTION: Swimming events are the sum of a gliding part and a swimming part. The gliding is used after the start and turns, and this phase typically corresponds to 10-25% of the total event time (Chatard et al., 1990). Taking this into account, one can notice that gliding is very important in swimming events and, therefore, its biomechanical study in order to make it more efficient is also very relevant.

The gliding can be studied experimentally, by using voluntary subjects gliding in a controlled manner in a swimming pool (using video or velocimetry, for instance), or by using Computational Fluid Dynamics (CFD). Although the experimental method gives "real" values it also presents some drawbacks, like usually imposing a heavy setup and also the fact that it is difficult to control all variables, like depth, attitude or intersegment positions of the swimmer. The CFD method does not have these limitations and its results are comparable to those obtained by the experimental method (Bixler & Riewald, 2002; Silva et al., 2005; Bixler et al., 2007; Vilas Boas et al., 2010).

This work aims to study the effects of the depth and velocity on the drag force experienced by a swimmer during gliding using the CFD method.

METHOD: A 3D scanning of a male swimmer in a streamlined position, with the arms extended above the head, was performed. The resultant 3D image was processed and converted to a format compatible with the CFD program Flow 3D (Flow Science, New Mexico - USA). The simulation domain is a parallelepiped with 11m length, 3m wide and 1m deep, simulating a swimming pool lane. The General Moving Object Model (part of Flow 3D) was used in the computations, with the water static and the swimmer model moving through it at constant longitudinal speed. The other swimmer velocity components were zero.

The model solves the Navier-Stokes equations, with turbulence governed by the Standard k-ε model (with parameters: $C\mu$ =0.09; C1=1.44; C2=1.92; σk=1.0; $\sigma\varepsilon$=1.3) – see for example Silva et al. (2005) for the equations used and the meaning of the symbols. The surface viscosity of the swimmer was 0.5; the initial turbulence was fixed at 1% with a scale of 0.1m. The water has a temperature of 28ºC, a density of 998.2 kg/m^3 and a viscosity of 0.001kg.m/s.

The simulations were performed for the swimmer speeds of U=1.5, 2.0 and 2.5 m/s, resulting in a value of about 10^6 for the Reynolds number. This value places the flow in a turbulent regime, as the critical value is about $5x10^5$ (Toussaint and Truijens, 2005). The swimmer model was "placed" horizontally on the computational domain. Considering the swimmer midline as the reference, the simulations were performed for the depths of 0m (swimmer half submerged), 0.1m (swimmer fully submerged, just below the water surface), 0.5m (middle of the swimming pool depth) and 0.9m (bottom of the simulation domain).

Since the swimmer was moving from the beginning of the simulation, the first two seconds of the simulation have a transient flow, with values fairly constant and stable afterwards. In this work the average results for the time interval between 2 and 3 seconds are presented.

RESULTS: The mean values for the drag force as function of speed and depth are shown in Table 1, for the time interval between 2 and 3 seconds.

Table 1. Mean values for the drag force

Depth (m)	Speed (m/s)	Drag force (N)	Depth (m)	Speed (m/s)	Drag force (N)
0	1.5	14.5	0.5	1.5	41.2
0	2.0	25.5	0.5	2.0	68.1
0	2.5	80.6	0.5	2.5	112.8
0.1	1.5	42.5	0.9	1.5	33.7
0.1	2.0	70.9	0.9	2.0	58.3
0.1	2.5	123.5	0.9	2.5	100.7

DISCUSSION: As shown in Table 1, the drag force for the half submerged swimmer is much lower than for the fully submerged swimmer at any depth, and for each speed. This arises from the fact that only half of the swimmer body is subjected to the water drag force, while the other half is subjected to the air drag force which is substantially lower due to its low density. For the fully submerged body simulations, the drag force decreases with increasing depth. This is probably due to a lower contribution of the wave drag force arising from the creation of waves at the air-water interface, as the disturbances caused by the swimmer at the air-water interface decrease with increasing depth. These results agree with the experimental results of Toussaint et al. (2002). As expected, the drag force increases with speed for all depths.

CONCLUSION: In this work we studied the effect of depth and speed on the drag force experienced by a swimmer moving in a streamlined position. Due to some limitations on the software, we were not able to separate the different components of the drag force (friction, pressure and wave), although from the analysis of the results we may conclude that the drag force due to waves decreases with increasing depth, while the other forms remain constant (except for 0m). These results may be useful for coaches and swimmers, as the gliding should be performed as deep as possible, although not so deep that the following rise to the surface for the swimming phase becomes too steep.

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KINEMATIC GAIT VARIABLES OF ELDERLY WOMEN WITH DIFFERENT LEVEL OF PHYSICAL ACTIVITY

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KEYWORDS: Kinematic gait variables, elderly women, level of physical activity.

INTRODUCTION: The aging of populations is an international phenomena caused by the decline of birthrate and the progress of medical science. According to WHO individuals older than 65 years in developed countries and older than 60 years in developing countries are considered elderly. The aging process is characterized by the decrease of muscle mass, strength and power, one of the most important reasons for the decrease of functional abilities and the increase of falling risk (Zhong et al., 2007). Nevertheless, physical activity may retard this process. The principal changes in kinematic gait variables are the decrease of gait velocity, stride length and single support time with an increase of double support time (McGibbon,2003). These changes in gait pattern observed in the elderly population may arise from functional declines of aging and may be even more significant in the absence of appropriate regular physical exercises. Considering the higher proportion of women in the elderly population (WHO, 2002), it is important to understand the changes in gait patterns of elderly women related to physical activity. Therefore, the purpose of this study was to identify the effects of different intensities of physical activity on the cinematic gait variables in older women.

METHODS: Forty four women between 65 and 75 years old participated in the study and were divided into three groups: 15 sedentary women, 15 physically active women and 14 athlete swimmers (table 1). The sedentary women did not practice any regular physical exercise, whereas the physically active group participated in a special physical exercise and sport program for elderly (5 times per week at least 30 min of predominant aerobic exercise or 3 times a week vigorous exercise for at least 20 min) according to the recommendation of the ACSM for promotion and maintenance of health of the elderly (NELSON et al, 2007). The athlete swimmers were veterans who still participated in international competitions and they were trained systematically. All subjects were free from any musculoskeletal or neurological disorder in lower limbs, spine and pelvis that could alter their gait pattern.

Table 1. Characteristics of the three investigated groups (mean and sd).

Variables	**Groups** Sedentary (n=15)	Active (n=15)	Athletes (n=14)
Age [years]	69.07 ± 2.84	67.87 ± 3.23	69.64 ± 3,81
Height [m]	1.56 ± 0.10	1.55 ± 0.05	1.58 ± 0.07
Body mass [kg]	61.80 ± 13.04	63.36 ±10.28	62.80 ± 9.49
BMI [kg/m^2]	25.33 ± 3.63	26.05 ± 2.96	25.21 ± 2.75

The GAITRite® walkway system (*MAP/CIR INK, Haverton, PA, USA*) was used to record the following variables over a distance of 5.6 m: Step and stride duration, absolute and normalized velocity which is defined as the ratio between absolute velocity and leg length [leg length/sec], single and double support time, oscillation time, step and stride length, step width, relation between step length and leg length. The participants walked at self-determined speed.
The retest reliability of the applied system has been investigated by different authors and the coefficients are reported to be between .92 and .95 (Menz et al., 2004).

The data were analyzed by one-way ANOVA and non-parametric Kruskal-Wallis tests for those variables that were not normally distributed. If significant differences were found between the groups, the *post hoc* Bonferroni test and the non-parametric Mann-Whitney U test ($p<0.05$) were performed.

RESULTS: The main results are shown in table 2. Significant differences were found between the sedentary women and athletes for gait velocity (p=0.004), step length (p=0.004) and stride length (p=0.003). Between the groups of physically active women and athletes significant differences were found for step length (p=0.004) and stride length (p=0.004). No significant differences were found between all groups for stride frequency, single support time, and double support time. For the investigated variables no significant differences between the sedentary and the physically active groups were identified.

Table 2. Kinematic gait variables of sedentary, active and athlete group (mean and sd).

Variables	Groups Sedentary (n=15)	Active (n=15)	Athletes (n=14)
Step length [cm]	64.94 ± 6.17	67.43 ± 5.41	72.49 ± 4.32
Stride length [cm]	130.04 ± 12.32	134.98 ± 10.86	145.18 ± 8.66
Stride frequency [min^{-1}]	117.81 ± 11.96	121.87 ± 10.28	125.61 ± 8.75
Normalized velocity [leg length/s]	1.52 ± 0.26	1.62 ± 0.18	1.80 ± 0.19
Single support time [s]	0.39 ± 0.04	0.40 ± 0.06	0.38 ± 0.03
Double support time [s]	0.26 ± 0.09	0.25 ± 0.12	0.22 ± 0.08

DISCUSSION: The results of the present study corroborate the findings of Lopopolo et al. (2006) that strength training or aerobic training at high intensity leads to higher gait velocities and corresponding changes of kinematic variables (e.g. step length, step frequency, support times) in the elderly. The activity of the physically active group might also cause physiological and kinematic adaptations (table 2). Nevertheless, these differences between the sedentary and physically active group were not significant. Even though a sample calculation had been performed, a greater number of subjects in each group may result in significant differences between the sedentary and physically active group too.

CONCLUSION: It was concluded that unspecific physical activities at higher intensities, such as configured in the athletes group, result in an increase of gait velocity, step length and stride length compared to sedentary females. The practice of even unspecific regular physical activity should be emphasized for older people in order to improve physical capacities, well-being, quality of life and reduction of accidents during gait.

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THE LEARNING PROCESS OF UNIFORMITY SKILLS FOR NOVICE ROWERS

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KEYWORDS: rowing, uniformity, oar, novice, learning

INTRODUCTION: In the crew events which row with a number of rowers, it is thought that the important technical element is the uniformity of crew how well rowers can synchronize timing of movement oars (Wing AM & Woodburn C, 1995; A Baudouin & D Hawkins, 2004). The highly uniformity skills also could make up for the total low power in the crew. In case of instruction for novice rowers, due to enhancement of uniformity skills, they may be able to feel the sensation of propulsive force of boat. Therefore, it is thought that this sensation would affect their interests in rowing. The purpose of this study was to identify the learning process of uniformity skills for novice rowers, and to obtain the basic data to instruct for novice rowers.

METHODS: In this program, participants were 200 students of Ishikawa National College of Technology in Japan. All students have never experienced for rowing. The contents at this program were instructions of rowing form on the Rowingergometers, and rowing boats for 20 minutes. Students divided into 5 groups (about 40 students per group). This program was held at Physical Education classes on a stated the day and time. 4 students get into rower seats (stroke, 3rd, 2nd, bow) and the instructor gets into cox seat on each boat. It was not specified that anyone takes any rower seats. The instructors engaged in giving students basic instruction (how to use oar, how to move their body). This program implemented 3 times/group (1 time per week).

The learning process of uniformity skill was estimated from the analysis of video images which was taken while participants were rowing boats. Parameters for estimating of the uniformity skills are time lag of CATCH-point (ms), time lag of FINISH-point (ms) and Rate (strokes/minute). Time lag of CATCH-point is the temporal difference between Stroke rower (who seated close to the stern of the boat) and other rowers when the blades are placed in the water. Time lag of FINISH-point is the temporal difference between Stroke rower and other rowers when the blades are lifted from the water. RATE is the number of strokes per minute by a crew.

In this research, Recreation boat, Macon blades (traditional U-shaped oar blade) and Rowingergometer (Concept2 TypeC) used to instruct for participants. Recreation boat is a shell with 4 rowers and cox, which makes it easier to learn basic rowing technique and less likely they will fall in water. Also, the filming while they were rowing boats was used Digital video camera (Sony, 30Hz).

The Freedman tests were used to analyze the changes in the mean of Time lag, which are reported as mean ± S.D. Spearman correlation coefficients were used to determine the strength of relationship between the parameters. The significance level was set at $p<0.05$ for all statistics.

RESULTS AND DISCUSSION: It was only 8 crews (32 students group) that could finish 3 sessions in this program. If Physical Education class was missed due to bad weather, they were excluded from this study.

The results of the mean in time lag of CATCH-point have been calculated (Figure 1). The means ± S.D. of this were resulted in 193.6± 79.2 ms (session1), 157.1± 58.0 ms (session2) and 155.8± 62.8 ms (session3). Time lag of CATCH-point at session2 and 3 was lower than at session 1. However, this did not result in gradual reduction of the time lag. Furthermore, it is thought that there was a wide spread of learning skills for novice rowers because the standard deviation were large for each session. As a result, it is thought that the magnitude of standard deviation was affected by the improvement of uniformity skills and the learning of rowing form.

Additionally, the result of relationship between time lag of CATCH-point and RATE was calculated (Figure 2). There was a significant correlation between time lag of CATCH-point and RATE (t=0.74, p<0.05). Therefore, it was suggested that it is effective for novice rower to set a low RATE when they learn to rowing uniformity skill.

In terms of FINISH-point, there was no significant change of the time lag and no significant correlation with other parameters.

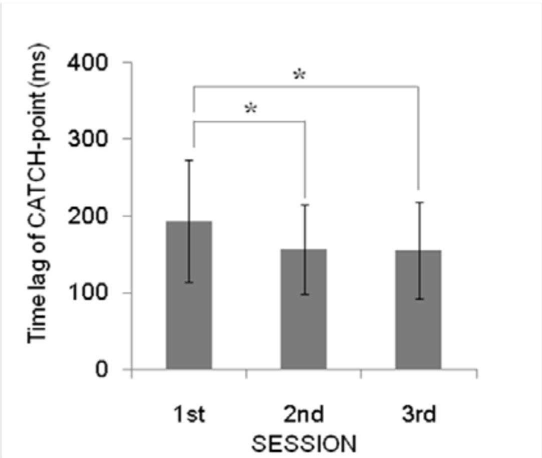

Figure 1. The mean of time lag of CATCH-point (ms) for each session

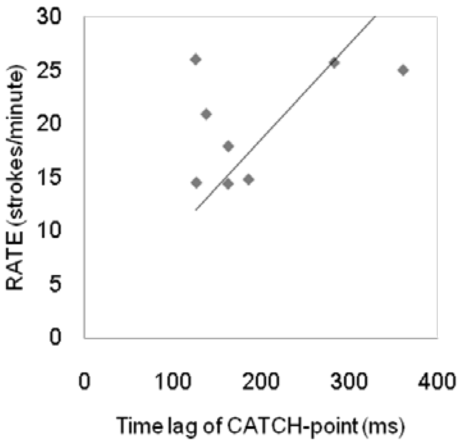

Figure 2. Correlation between the mean of time lag of CATCH-point (ms) and RATE (strokes/minute)

CONCLUSION: As these results of this program, it could identify the learning process of uniformity skill for the novice rowers. It is important for novice rower to set the low rate and to pay attention to CATCH-point in order to enhance the uniformity skills. We will need the longer term research in the future.

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KINETIC EFFECT OF A FOUR-STEP AND STEP-CLOSE APPROACH IN A VOLLEYBALL SPIKE JUMP FOR FEMALE ATHLETES

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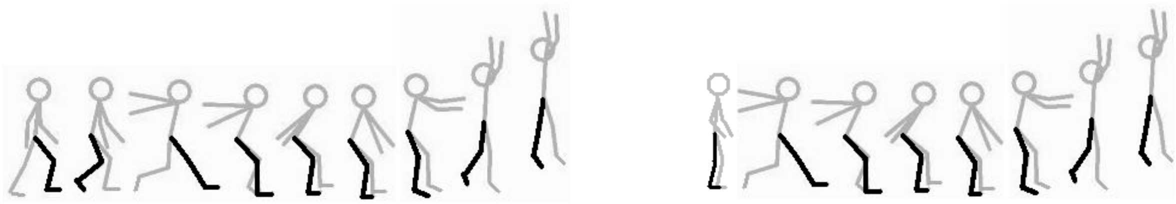
The purpose of the present study was to investigate the kinetic difference between two different volleyball spike jump techniques: a complete four-step approach and step-close approach. Five female collegiate volleyball players (age: 20.40 ± 1.85, height: 1.80 ± 0.02 m, body weight: 71.71 ± 4.18 kg) who play the middle hitter position were recruited. Each participant performed ten jumps for both four-step and step-close approaches and takeoff from two Kistler force platforms. Results indicated that there is no significant difference (P = .18) of vertical propulsive impulse between the two types of jump. The anterior-posterior (AP) net impulse of the four-step approach was significantly greater than a step-close approach (P < .01). Finally, the contact duration of propulsive phase for step-close technique is significantly greater than four-step approach technique (P < .05).

KEYWORDS: approach, impulse, kinetics, volleyball spike jump.

INTRODUCTION: The International Volleyball Federation (FIVB) defined attack hit as "*All actions which direct the ball towards the opponent, with the exception of service and block*" (FIVB, 2008). The volleyball attack hit is an important offensive tool which dominates the result of a competition. One type of the attack hit involves maximum jump height with approach. Jump height of the hit provides the advantage of attack angle and time in the air (Abendroth-Smith & Kras, 1999). The mechanism of jumping has been investigated from many different perspectives which incudes muscle mechanics and segmental kinematics and kinetics (i.e., Dapena & Chung, 1988; Moran & Wallace; Vint & Hinrichs, 1996; Wagner, Tilp, Duvillard, & Mueller, 2009). Kayambashi (1977) indicated that the number of approach steps resulted in different jump height for male volleyball players. Hsieh and Christiansen (under review) indicated that there is no significant relationship between approach velocity and jump height in women volleyball players. By comparing two different types of jump, step-close and hop jump, there was no significant difference at jump height or vertical impulse between the two styles (Coutts, 1982; Gutiẻrrez-Davila, Campos, & Navarro, 2009). However, the vertical impulse is enhanced as the last step length increases for male volleyball players (Liu, Huang, & Huang, 2001). Therefore, these studies showed that different approach techniques have different effects on jump height for men and women volleyball players.

In a regular volleyball match, the middle hitter has to play an active role by running a "quick" or "slide" in front of or behind the setter regardless if the setter is going to set the ball to him or her. In many situations, especially during a rally, the middle hitter has little or no time to pull back far enough to perform a complete three- or four-step approach after a block; instead, he/she has to take a step-close approach to jump (Figure 1). Thus far, no research has examined the kinetic comparisons between a four-step and step-close approach jump in female volleyball players. Therefore, the purpose of the present study was to investigate the



Four-step approach Step-close approach

Figure 1. Figures above show a complete four-step approach and step-close jump (Note: solid black line represents right lower extremities).

propulsive vertical and horizontal impulse (AP impulse) between a four-step and step-close approach jump in female volleyball players. Furthermore, the results may provide suggestions to the middle hitter for different techniques to improve jumping skills.

METHOD: A total of five female collegiate volleyball players (age: 19.40 ± 1.85, height: 1.80 ± 0.02 m, body weight: 71.71 ± 4.18 kg) were recruited. All policies and procedures for use of human subjects were followed and approved by the local Institutional Review Board. All participants had at least seven years of experience playing competitive volleyball and their current position on the court is middle hitter.

Each participant was requested to take five minutes of warm up with jogging and stretching before the data collection. After warm up, each subject was required to practice jumping on the two Kistler force platforms (Model 9286; 600 Hz) in order to mark the starting position for either a four-step approach spike jump or step-close approach spike jump. Kistler Bioware® software was used to analyze 5 seconds of force data. During the data collection, each participant was required to perform ten maximum volleyball spike jumps using four-step approach and another ten using step-close style to jump as high as possible. A trial was excluded when the subject's feet failed to make full contact with both force platforms during takeoff phase. Thirty seconds and two-minute breaks were provided between trials and after the last trial, respectively.

A standard t-test was performed to compare the difference of propulsive vertical and horizontal impulses between a complete and step-close approach spike jump. The time between takeoff and landing from the GRF data was used to estimate the takeoff velocity (v = t/2*g) and the vertical propulsive impulse was calculated with body mass (vertical propulsive impulse = m*v) (Liu et al., 2001). Horizontal impulse was determined by using formula: $\int_{T_{initial}}^{T_{End\ of\ Braking\ Phase}} (F_{AP\ GRF}) \times \Delta time$, where time was the duration of positive AP force (braking phase). The duration of contact was from the beginning of the first foot impact to takeoff. To control for both type I and II errors, Holm's correction formula was utilized to calculate new adjusted critical *P*-value $= \alpha/(n-i+1)$, where *n* is the total number of comparisons and *i* is the order of comparison (Knudson, 2009; Lundbrook, 1998). Each observed *P*-value was compared to new adjusted critical *P*-value according to the equation provided.

RESULTS: Results showed that there was no significant difference of propulsive vertical impulse between the two types of approach techniques (*P* = .183). The complete four-step approach had significant greater horizontal impulse with *P*-value less than .001. Finally, the time for the contact phase indicated that the four-step approach had significant shorter duration with *P*-value of .03. Figure 2 represents a sample force-time graphs of vertical and AP GRF for both types of approach technique. The beginning of the foot contact for both type of jumps were matched in order to show the difference.

Table 1. Vertical and horizontal impulse and contact duration for both approach techniques

Techniques	Vertical Impulse (J)	Horizontal Impulse (J) *	Duration (s) *
Four-step approach	226.60 ± 26.65	136.89 ± 25.38	0.38 ± .05
Step-close approach	217.57 ± 24.81	104.68 ± 33.69	0.45 ± .15

Note: * indicates significant difference with *P*-value less than new adjusted critical *P*-value.

DISCUSSION: This study examined the effect of a complete four-step and step-close approach in volleyball spike jump for female players. Kayambashi (1977) investigated national team volleyball players and found that the number of steps was not correlated to the jump height in female athletes which supports the finding of the present study that there is no significant difference (*P* = .183) of the propulsive vertical impulse between four-step and step-close approach jump for this group of female players. The other finding in this study showed that the four-step approach had greater braking horizontal impulse than the step-close approach which also implies that a complete approach had greater horizontal momentum (velocity). Studies have found that horizontal velocity is a crucial variable toward jump height when male participants were recruited for this special type of jumping technique (Liu et al., 2001; Wagner et al., 2009). However, in the current study using a group of female volleyball players, the results seem to contradict that finding.

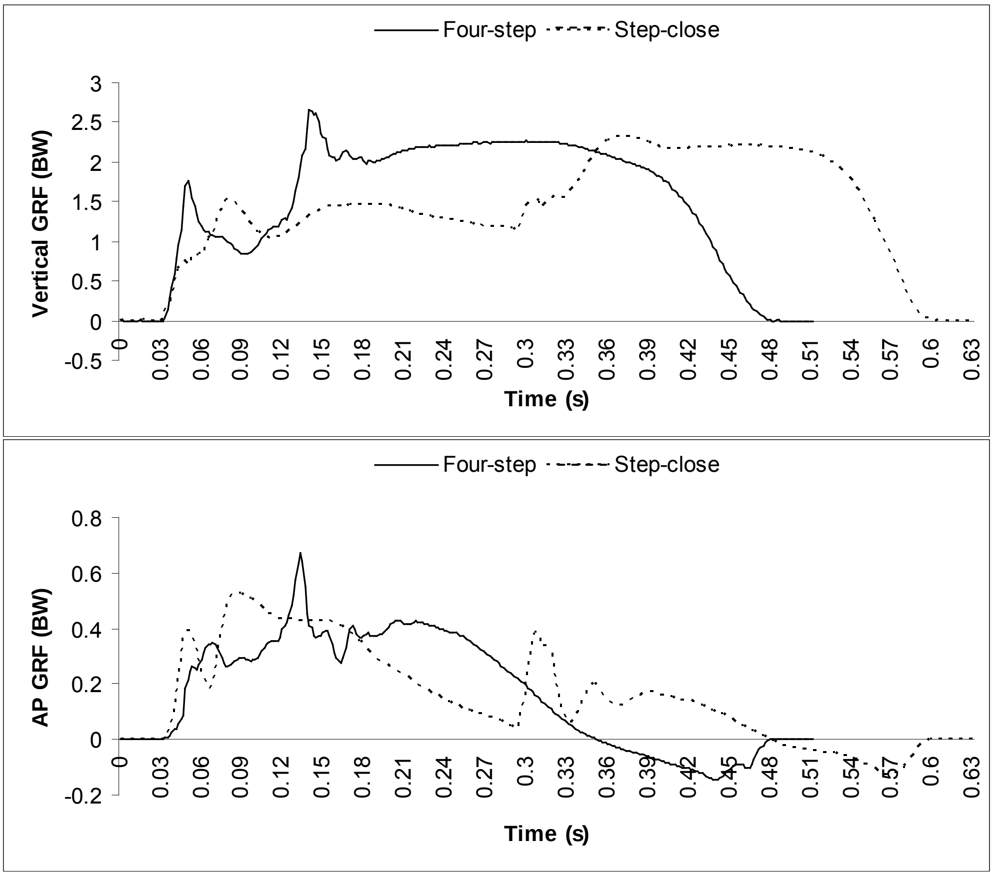


Figure . Sample vertical and AP GRF from one trial of a subject.

Results also showed that the four-step technique had significant shorter duration of contact phase which may result from greater horizontal momentum that speeds up the angular motion of the COM pivot over the supported legs to takeoff in a shorter amount of time. In addition, with greater horizontal momentum the completion of both feet impact was earlier than the step-close technique. On the other hand, the rate of vertical force development on the force-time graph was found to be slower in step-close technique due to the plant of both feet are further apart on timing when compare to four-step approach (Figure 2). This, coupled with the longer duration of contact phase, compromised the discrepancies of vertical impulse exertion at beginning of the contact phase for the step-close technique.

For a regular volleyball attack jump, studies have found that horizontal motion is minimized during the takeoff phase (i.e., Prsala, 1982). Additionally, Chen and Huang (2008) found that the back row attack had greater jump height when compared to front row attack for elite female volleyball players due to greater horizontal velocity at takeoff. This implies that when horizontal displacement is allowed after takeoff, the horizontal momentum from approach may have efficiently contributed to the jump height. Therefore, with significant different horizontal impulses, it may have represented different direction of the resultant GRF at beginning of the contact phase. This could explain how the transition of horizontal and vertical momentum was made. However, without further kinematic data such as the radial motion of the COM, it would be difficult to determine the mechanism of these two different approach techniques and how this group of athletes maintained similar results of vertical impulse. Another limitation includes a sampling frequency set at 600 Hz which may have slightly underestimated the jump height by less than 1% (Street, McMillan, Board, Rasmussen, & Heneghan, 2001). Finally, all the subjects were required to perform the jumping techniques in the laboratory setting which may have influenced the jumping performance when compared to performing on the volleyball court, such as minimum horizontal displacement after takeoff.

CONCLUSIONS: The present study showed that step-close approach can create the similar amount of propulsive vertical impulse as the four-step approach spike jump which indicated the jump height is similar for both techniques. The AP impulse during braking phase showed that this group of performers had greater horizontal velocity due to the four-step approach technique. However, this horizontal momentum did not contribute to the vertical jump as other studies' findings when male subjects were recruited. Finally, the step-close technique has a longer period of contact phase which indicated that the average vertical force exertion may be different between the two techniques. Therefore, for this group of female volleyball players who play the middle block position may have a different jumping mechanism for both types of jumps to maintain similar performance results.

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A METHOD TO ANALYZE SOCCER OFFENSIVE SEQUENCES

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KEYWORDS: Soccer, attack velocity, offensive sequences.

INTRODUCTION: There is a range of possibilities to analyze the soccer game in relation to physical-technical-tactical aspects. The importance of going toward the goal rapidly since the recovery of possession was defended by Grehaigne et al (1996). The main technique to analyze soccer attack speed in literature was presented by Yue et al (2008). The aim of this study was to propose a method to analyze offensive sequences in soccer based on goal progression velocity (GPV) and goal progression indicator (GPI).

METHODS: One game from Brazilian first division championship was recorded with four digital cameras (30 Hz). Players' positions during the match were obtained using Dvideo software as in Figueroa et al (2006a, 2006b). Technical actions were also registered using Dvideo as in Moura (2006). Ball position was considered as the position of the player just when the action was executed. All offensive sequences from both teams that ended in shots and with at least two actions were considered into the analysis (n=29). Two of them (which ended in goals) were detailed.
In each attack, for every pair of sequential actions (i and $i+1$), the GPV was calculated by: $\overrightarrow{\mathrm{GPV}} = \left(\overrightarrow{\mathrm{vel}} \cdot \overrightarrow{\mathrm{tar}}\right) \overrightarrow{\mathrm{tar}}/|\mathrm{tar}|^2$, where $\overrightarrow{\mathrm{vel}}$ is a vector with direction defined by vector $\overrightarrow{bt}$ (figure 1) and modulus equal to the ball average velocity between these actions; $\overrightarrow{\mathrm{tar}}$ is the vector defined by the center point between the same actions and the center point of the goal. As a result, $\overrightarrow{\mathrm{GPV}}$ is the projection of $\overrightarrow{\mathrm{vel}}$ over $\overrightarrow{tar}$. The modulus of $\overrightarrow{\mathrm{GPV}}$ represents the velocity and the signal – positive or negative – of scalar multiplication $\overrightarrow{vel} \cdot \overrightarrow{tar}$ indicates progression or digression in relation to the goal. Figure 1 shows an example.
The GPI was calculated for the same actions by: $\overrightarrow{\mathrm{GPI}} = \left(\left(\overrightarrow{\mathrm{bt}}/2 \cdot \overrightarrow{\mathrm{tar}}\right) \overrightarrow{\mathrm{tar}}/|\mathrm{tar}|^2\right)/\overrightarrow{\mathrm{tar}}$, where $\overrightarrow{\mathrm{bt}}$ is the vector defined by actions i and $i+1$. Consequently, $\overrightarrow{\mathrm{GPI}}$ is the projection of $\overrightarrow{\mathrm{bt}}/2$ over $\overrightarrow{tar}$ normalized by $\overrightarrow{tar}$. The signal – positive or negative – is given by the scalar multiplication$\left(\overrightarrow{bt}/2\right) \cdot \overrightarrow{tar}$. Therefore, GPI can exist between -1 and 1, where 1 is the biggest progression to goal possible and -1 is the biggest digression.

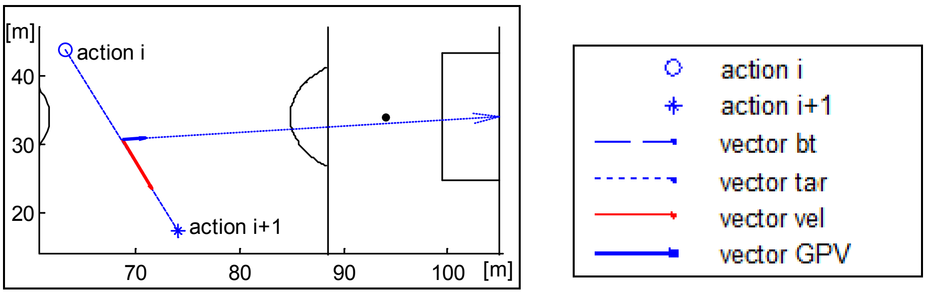

Figure 1. Examples of $\overrightarrow{GPV}$, $\overrightarrow{vel}$, $\overrightarrow{bt}$ and $\overrightarrow{tar}$. $\overrightarrow{GPV}$ is the projection of $\overrightarrow{vel}$ over $\overrightarrow{tar}$.

As a result, GPV shows the attack velocity while GPI is a relation among the goal progression and the goal distance. GPV and GPI maximum and minimum values will presented in average and standard deviation. Results from two sequences ended in goals will be detailed.

RESULTS: The average of GPV and GPI minimum values for all offensive sequences ended in shots in both halves were 2.5 m/s (±5.5) and 0.21 (±0.17), respectively. Also, average of maximum values of GPV and GPI were 10.4 m/s (±5.7) and 0.31 (±0.33), respectively. Each GPV and GPI values from two goals on second half are presented in figures 2 and 3. In these cases, GPV and GPI maximum values were 13.2 m/s and 0.49 for goal 1 and 11.3 m/s and 0.41 for goal 2, respectively.

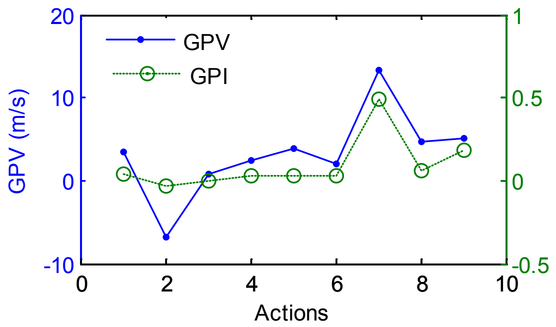

Figure 2. GPV and GPI from all goal 1 actions.

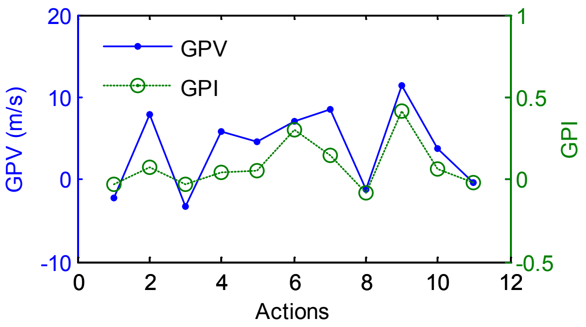

Figure 3. GPV and GPI from all goal 2 actions.

DISCUSSION: An advantage of the method proposed is that it permits the analyses of the influence of each action to attack velocity and goal progression, and not only the average attack speed. Goal 1, for example, presented higher values of GPV concentrated in the last 3 actions, while goal 2 showed alternated high and low values of GPV, showing different characteristics between these sequences. Also, the use of GPI allows the analyses with emphasis on the actions in which the ball gets close to the target. Goal 1, for example, presented only one high value of GPI (action 7), while goal 2 presented two peaks (actions 6 and 9). GPV peaks do not necessary indicate goal proximity and combined with GPI may be a sign of which actions were the most important to the sequence success.

CONCLUSIONS: Although there are various manners to reach the goal on soccer, finding more effective ways to score is very important to all teams. The GPV and GPI average and standard deviation values from every offensive sequences ended in shots from one game were presented. The method proposed on this paper showed how offensive sequences can be analyzed considering the progression to the goal and attack velocity. Studies with a greater number of matches are necessary from now on.

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FRONT FOOT SLIDE VARIABILITY AND ITS RELATION TO TENPIN BOWLING PERFORMANCE

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KEYWORDS: Tenpin Bowling, Variability, Foot Slide

INTRODUCTION: In tenpin bowling, bowlers try to knock down as many pins as possible with the allotted number of tries. In the modern power game, they achieve this by generating a lot of momentum using heavy balls released accurately at great velocities (Strickland, 1996). It must be done consistently over many tries. Although accuracy of the front foot slide during the delivery phase seems less relevant compared to the accuracy of the ball release, its consistency is still paramount. The ability to slide the front foot consistently presumably enables the bowler to have a predictable stable base to deliver the ball more accurately. Variability in performing sports skills has been studied in various disciplines such as javelin and basketball (Bartlett, 2008), but no published data is available on tenpin bowling. In fact, published work on tenpin bowling is rather scarce; the only recent study was by Chu and colleagues (2002) which compared a number of kinematic variables in the sagittal plane between male and female bowlers. The purpose of this study was to look at how variability of the front foot slide was related to average bowling score (B_{ave}) and ball release velocity (BR_{vel}). In addition, the variability of the foot kinematics between elite and amateur bowlers was also compared.

METHODS: Participants were assigned into two groups based on their B_{ave} over 3 tournaments, with those averaging above 200 pin falls placed in the elite group. There were 18 elite (M=10, F=8; B_{ave} 213.2±6.80) and 12 amateur bowlers (M=7, F=5; B_{ave} 181.3±9.36). Foot kinematic data were derived from Kwon3D motion analysis system, while BR_{vel} was measured by timing gates. Four Basler (100Hz) cameras were used for motion capture, which was carried out at the bowling alley. The participants aimed for a strike at each delivery, meaning that pins were reset after each trial irrespective of whether there were any pins left standing. There were 7 trials and bowlers were instructed to use similar delivery methods for every trial. However, only trials 3 to 6 were used in the analysis. Reflective markers (15mm) were placed at the front tip of the bowling shoe and under the heel counter, the midpoint between the two markers was taken as the foot position. Time series data were normalized from top of back swing (TBS) to the top of follow through (TOF). The following discrete and continuous Foot Position (FP) variables were determined; Anterio-Posterior (AP) and Medio-Lateral (ML) position, AP velocity and acceleration. The mean Standard Deviation (SD) of the trials were used as the variability indicator of the front foot slide, while mean peak values were used for velocity and acceleration variables. Independent samples t-test and Pearson Product Moment Correlation were used to compare groups and look at relationships, respectively. Significance level was set at $p<0.05$.

RESULTS: The results relating to the group comparisons and relationships of the front foot slide variability is summarised in Table 1. Meanwhile, in terms of the relationship between BR_{vel} with foot velocity and acceleration; BR_{vel} was significantly correlated with peak foot velocity (r = 0.52), peak acceleration (r = 0.39) and peak deceleration (r = -0.48). Unsurprisingly, BR_{vel} was significantly correlated to the B_{ave} (r = 0.59).

Table 1

Variable	Correlation: Mean SD	Correlation: Coefficient	Group Comparison: Group	Group Comparison: Mean SD	Group Comparison: t-value
AP foot position at FS	0.065	▲$r = 0.46$	Elite	0.736	$t_{28} = 2.48$*
			Amateur	0.525	
AP foot position averaged	0.415	+$r = 0.53$	Elite	0.132	$t_{28} = -0.96$
			Amateur	0.150	
ML foot position at FS	0.015	▲$r = -0.46$	Elite	0.013	$t_{28} = -2.88$*
			Amateur	0.019	
ML foot position averaged	0.016	▲$r = -0.40$	Elite	0.014	$t_{28} = -2.33$*
			Amateur	0.019	

▲Significant correlation to bowling average
+Significant correlation to ball release velocity
*Significant difference

DISCUSSION: In terms of B_{ave}, it seems that the consistency of left to right position of the front foot plays a significant role, with results showing that the lower variability in ML position at FS prior to the slide correlated with better bowling scores. This is supported by the fact that those who had followed a more consistent ML foot path throughout delivery had also shown better B_{ave}. Furthermore, the elite group were more consistent in ML position at FS and had less variability of ML position throughout. Interestingly though, higher variability in AP foot position at FS seem to relate to better B_{ave}. This could possibly be explained by the results of the correlation figures of BR_{vel}. Higher variability of AP foot position throughout delivery was related to higher BR_{vel}. But, taking into consideration that higher BR_{vel} also relates to higher peak foot velocity, higher acceleration and deceleration of the foot, the results are somewhat expected. As it is necessary for bowlers to stop their slide before the foul line, it is possible that a bowler moving at greater speeds will invariably constantly need to adjust their AP foot position in order not to pass the line. This is commonly referred to as the speed-accuracy trade-off (Schmidt et. al., 1979); trying to perform a skill quickly tends to lead to less accuracy and less consistency. Even though moving at higher velocities can lead to higher AP variability, good bowlers still seem keen to achieve higher momentum, preferring to just concentrate in minimising ML variability. The importance of BR_{vel} is highlighted by its significant correlation to better B_{ave} and supports the notion that modern bowling is very much a power game.

CONCLUSION: Accuracy and consistency have been acknowledged as important factors in tenpin bowling, but its specific application to training is rather limited. Results of this study indicate that consistency of side to side foot path during slide and foot placement at the start of the slide is important to achieving better bowling scores. In terms of practice, in addition to the already available markers at the front end of lanes, coaches may want to consider leaving markers or drawing explicit lines on the lanes during training to help achieve more consistent foot positions during the slide. Consequently, front to back position of the sliding foot should be expected to vary as the bowlers try to adjust their body segments to stop before the foul line whilst trying to attain greater ball release velocities.

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THE EFFECT OF UPPER EXTREMITY USAGE ON TRANSFER OF ANGULAR MOMENTUM DURING SOCCER INSTEP KICK MOTION

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KEYWORDS: soccer, remote term of angular momentum, instep kick, upper extremity

INTRODUCTION: The instep soccer kick has been subject to the majority of biomechanical analysis and research. However, very few studies have investigated the segmental contributions of the upper limb to the generation of angular momentum during the instep kick. Shan & Westerhoff (2005) reported that upper body movement contributes notably to skill effectiveness during soccer instep kick. Since the arms will likely experience rapid changes in position during the kicking, movements of the arms may be used to regulate excessive angular momentum on the lower limb. Therefore, the accuracy and ball speed for the instep kick can be gained from proper use of the upper limb. The purpose of this study was to investigate the effect of the upper extremity usage on transfer of angular momentum during soccer instep kick motion.

METHOD: Ten male soccer players (20.5±1.5 yrs, 176.4±2.9 cm, 663.1±64.7 N), each with at least 6 years kicking experience, were recruited as the participants. Kinematic data from four digital camcorders (Sony DCR-HC48, 60 fields/s) were collected while subjects were asked to kick the ball for four conditions (WA: With arms, WOLA: Without Left arm, WORA: Without Right arm, WOA: Without arms) in random order. For instep kick without arm movement, the subject's hands were folded across the chest throughout the kick. Twenty two reflective makers and four external stick markers were attached to the body. The markers were digitized manually. The remote term of angular momentum (RAM) was computed for each trial. The RAM represents the moment of linear momentum relative to the system center of mass. For each dependent variable, a one-way analysis of variance (ANOVA) with repeated measures was performed to test if significant difference existed among different four conditions (p<.05).

RESULTS: For all participants, the medio-lateral RAMs were generally greater for WA. The medio-lateral RAM of the right thigh at the LC (Left foot contact) in WA and WORA were significantly greater than the corresponding value in WOA (table 1).

Table 1. Medio-lateral remote angular momentum at the LC (unit: kg·m^2/sec)

	WA	WOA	WOLA	WORA
Trunk	-15.91(3.16)*	-15.08(2.67)$^{\partial}$	-16.09(2.82)$^{\partial\oint}$	-14.26(2.46)$^{*\oint}$
R. Thigh	11.80(2.06)*	10.04(1.83)$^{*\partial}$	10.70(2.13)*	10.94(1.81)$^{*\partial}$
R. Shank	11.68(2.91)*	9.22(2.83)$^{*\partial}$	10.38(2.88)*	10.45(2.26)$^{*\partial}$
R. Foot	2.37(1.39)*	1.09(1.48)*	1.74(1.40)	1.65(1.16)

Note. *significant difference between WA and WOLA, WORA, WOA, $^{\partial}$significant difference between WOA and WOLA, WORA, $^{\oint}$significant difference between WOLA and WORA at p<.05, Standard deviation in parentheses.

At the BI (Ball impact), the medio-lateral RAM of the right foot in WA was also significantly increased (table 2). It is interesting to note that the anterior-posterior RAM of the trunk at the BI in WA and WORA were significantly smaller when compared to WOA and WOLA (table 3).

Table 2. Medio-lateral remote angular momentum at the BI (unit: kg·m^2/sec)

	WA	WOA	WOLA	WORA
Trunk	-15.27(3.24)	-14.53(2.96)	-14.89(3.23)	-14.25(2.71)
R. Thigh	5.90(2.10)	5.04(1.77)$^{\partial}$	5.85(2.03)$^{\partial}$	5.31(1.73)
R. Shank	15.01(3.21)*	13.26(2.57)$^{*\partial}$	14.01(2.90)$^{*\partial}$	14.01(2.78)$^{*\partial}$
R. Foot	8.58(1.09)*	8.02(0.93)*	8.09(1.17)*	8.26(1.08)*

Table 3. Antero-posterior remote angular momentum at the BI (unit: kg·m^2/sec)

	WA	WOA	WOLA	WORA
Trunk	0.61(2.01)*	1.38(1.97)$^{\partial}$	1.96(2.02)$^{*\oint}$	0.04(1.97)$^{\partial\oint}$
R. Thigh	-0.39(0.74)	0.17(1.14)	-0.40(0.69)	-0.08(0.85)
R. Shank	1.39(1.14)*	0.91(1.47)	0.87(0.66)*	1.32(1.50)
R. Foot	0.83(0.57)*	0.34(0.77)*	0.47(0.33)*	0.63(0.75)

DISCUSSION: It has been postulated angular momentums in the lower limb and the trunk were directly influenced by the arm movement. The results of this study showed that WOA, WOLA, and WORA clearly had a smaller RAMs than WA. This indicates that the increase in RAMs was thought by some to be the result of the arm's motion. Bezodis et al. (2007) reported that movement of the non-kicking side arm helps to prevent over rotation of the whole body about the vertical axis. Shan & Westerhoff (2005) showed that a skilled player will use arm extension on the non-kick side to make a tension arc at the beginning of the kick. Flexion and adduction of the non-kicking-side arm were found to be widely used by skilled kickers during the kicking. These movements were suggests as one cause of the increased ball speeds exhibited by skilled kickers (Shan & Westerhoff, 2005). Rotational countermovement of the arm was found to be widely used by all participants in this study. It is likely that WA exhibited a greater potential to alter their RAM profiles about each axis, which may be beneficial for performance. However, in case of WORA, it seems that the angular momentum possessed by the left arm can not be used to control excessive angular momentum generated by the lower limb.

CONCLUSION: The present study showed that RAMs during instep kicking were greater in WA compared with WOA, WOLA, and WORA. It is more likely that the upper and lower limb segments interact to maintain a balanced during instep kick. Thus, it was concluded that upper limb segments played an important role in regulating the angular momentum generated by the lower limb. It was also suggested that difference in kick biomechanics depend on the skill level of the players. This has led to suggestions of the need for further analyses using different participant population.

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EFFECT OF THE VELOCITY OF THE CENTER OF MASS IN PERFORMING THE BASKET WITH HALF TURN TO HANDSTAND ON PARALLEL BARS

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KEYWORDS: gymnastics, parallel bars, basket, turn.

INTRODUCTION: The basket with half turn to handstand on parallel bars (Fig. 1) is an important skill in men's artistic gymnastics. Techniques of the turn in performing this skill are classified into two types: early turn and late turn. With the early turn, the gymnast can improve the difficulty value by increasing the angle of the turn. Peak forward velocity of the center of mass in the early turn has been shown to be greater than in the late turn during forward swing, but peak upward velocity differs little between the two types of turn (Yamada et al., 2009). Those results spawned two hypotheses concerning the limit to which kinetic energy can be increased during the early turn. (1) Intrinsic characteristics of the technique restrict how much kinetic energy can be incorporated into the early turn. (2) The gymnast's muscular ability to generate shoulder flexion torque and hip extension torque limits how much kinetic energy can be produced in the early turn. To explore these hypotheses, it might be useful to compare the basket with half turn to one without a half turn in arriving at handstand. The purpose of this study was to compare the effect of velocity of the center of mass on turn technique in a basket with half turn to handstand versus a basket without turn to handstand.

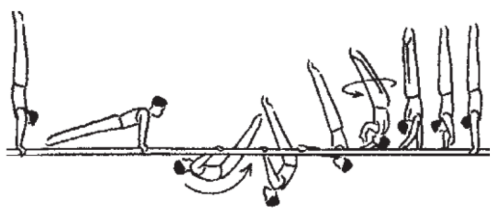

Figure 1. The basket with half turn to handstand (from FIG, 2009).

METHOD: Four senior male gymnasts competing nationally were asked to perform the basket to handstand and the basket with half turn to handstand. The gymnasts repeated these maneuvers until they, along with a coach with a license to judge, agreed that performance was satisfactory. All of these performances were videotaped using two digital video cameras (60 Hz)(Sony, DCR-VX1000 and DCR-TRV900), one from a lateral view and the other from a diagonal view in front. Twenty-two body landmarks (right and left third MPs, wrists, elbows, shoulders, great toes, heels, ankles, knees, and hips, and in the midline the vertex, midpoint between tragions, suprasternale, and lower end of the thorax) were digitized (DKH, Frame-DIAS IV). Three dimensional coordinates were synchronized and reconstructed using the method of Yeadon and King (1999). The coordinates were smoothed with a fourth order Butterworth digital filter with cut-off frequencies ranging from 3.4 to 5.4 Hz, and the center of mass of each segment and of the whole body were estimated using the body segment inertia parameters of a Japanese athlete model (Ae, 1996).

RESULTS: Two subjects were classified as using the early turn and the remaining two subjects the late turn, according to observations by the coach. Table 1 shows peak horizontal (forward) and vertical (upward) velocities of the center of mass in the basket to handstand and the basket with half turn to handstand. Peak horizontal velocity of the center of mass appears at around the lowest point of the swing and then vertical velocity reaches peak value. Peak horizontal velocity of the center of mass was similar between using and not using the turn, whether early or late. Whereas in the subjects using the early turn peak vertical velocity in the basket was greater in the performance without the turn than with the

Table 1. Peak horizontal (forward) and vertical (upward) velocities of the center of mass in performing basket without and with half turn to handstand.

Subject	Type of Turn	Peak Horizontal Velocity Without Turn (m/s)	Peak Horizontal Velocity With Turn (m/s)	Peak Vertical Velocity Without Turn (m/s)	Peak Vertical Velocity With Turn (m/s)
1	Early	3.37	3.52	3.73	3.61
2	Early	3.33	3.38	3.94	3.72
3	Late	3.17	3.10	3.56	3.65
4	Late	3.19	3.29	3.83	3.92

turn, in the subjects using the late turn peak vertical velocity in the basket was a little greater with the turn than without the turn.

DISCUSSION: Horizontal and vertical velocities in the late turn were not systematically influenced by presence or absence of the turn itself. The late turn consisted of a basket to handstand on one rail and a half turn backward in a handstand. The belatedness of the turn appears to have made its effect on what happened during the upward phase of the swing relatively small. In the early turn, on the other hand, vertical velocity was smaller in the basket with half turn to handstand than without that turn, possibly because shoulder flexion and hip extension during the upward phase of the swing might have been impeded by the turning movement, given its start from within the basket swing itself. Vertical velocity at bar release is important for performing the basket to handstand mount (Takei and Dunn, 1996). Vertical velocity at bar release is considered to be important, whether or not a turn intervenes in arrival to the handstand. In the early turn, peak vertical velocity could not be increased with the turning movement, so the gymnast appears to have had to compensate for the limitation in vertical velocity by augmenting horizontal velocity. The results here suggest that large horizontal velocity through the downward swing from the handstand position was important to achieve in the technique of the early turn.

CONCLUSION: The turn techniques were classified into early and late types. In the subjects using the early turn peak vertical velocity in the basket was greater in the performance without the turn than with the turn, so the hypothesis that intrinsic characteristics of the technique limit kinetic energy appears more plausible than the hypothesis about the gymnast's muscular ability. Horizontal velocity was shown to be important for executing the early turn technique because shoulder flexion and hip extension were rendered difficult by the turning movement. If a gymnast uses a large horizontal velocity in the basket to handstand, the early turn would be the recommended technique, whereas if horizontal velocity is not so great, either the late turn should be adopted or the gymnast should improve movement during the downward swing to augment horizontal velocity.

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INFLUENTIAL LITERATURE IN APPLIED SPORTS BIOMECHANICS

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This study documented the perception of prestige of applied sports biomechanics journals, as well as influential articles and books. Recent ISBS members were surveyed to rate the quality/prestige of 35 journals. Descriptive statistics of ratings were calculated for respondents and correlated with the 2008 impact factor (IF) reported in the *Journal Citation Reports*. Mean ratings showed that international perception of influential journals were weakly (r = 0.48) correlated with the IF. These results confirm previous studies that the IF is a poor index for evaluating the influence of journals publishing applied sports biomechanics research, and there was considerable diversity among the respondent's nominations of the most influential books and articles in the field.

KEY WORDS: bibliometrics, book, citation, impact factor.

INTRODUCTION: Much of the world scientific community uses the *Journal Citation Reports* and the impact factor (IF) statistic to rate journals. There have been, however, many articles reporting several limitations of the IF (Bollen et al. 2006; Frank, 2003; Seglen, 1997). One problem of the IF that adversely affects sports biomechanics is a bias against small disciplines (Frank, 2003; Seglen, 1997; Stanzer 1995).

This problem has been explored by several studies in biomechanics. Knudson and Chow (2008) reported that ratings of journal prestige by American Society of Biomechanics (ASB) members were weakly correlated (r = 0.35) with the IF. This weak association was confirmed (r = 0.34) by a study of International Society of Biomechanics in Sports (ISBS) member ratings of journals and the IF (Knudson and Ostarello, 2008). Knudson and Chow (2008) also reported journal ratings were different relative to the ASB interest area of the respondents. For example, "ergonomics & human factors" members rated *Ergonomics* in the top three, while the other four interest areas did not have this journal in the top 13 of 62 journals. Exercise & Sport Sciences members rated *Sports Biomechanics* 11th but other interest areas did not have it in the top 20. Knudson (2007) reported that journal ratings by exercise & sport sciences members of ASB were not significantly correlated with the IF.

It is likely that scholars in an applied discipline like sports biomechanics are disadvantaged by the widespread use of the IF. Theoretical and multidisciplinary journals have nominally higher IF values than clinical or applied journals. Given that sports biomechanics serials are either weakly or not correlated with the IF, it is important for applied disciplines like sports biomechanics to document the peer ratings of important journals and literature in their fields. The purpose of this study was to document the perception of prestigious or influential applied sports biomechanics journals by ISBS members. The study also examined the articles and books that were considered influential and highly regarded in the field. It was hypothesized that ratings of journals and highly-rated articles would be poorly associated with journal 2008 IF.

METHOD: An invitation to participate in a survey listing 35 journals (Table 1) was emailed to the recent members (n = 916) of ISBS. Journals publishing applied sport science research that were recently rated in previous studies (Knudson 2007; Knudson & Ostarello, 2008) were included in the survey. Two more emails reminded members of the opportunity to participate in the study. Respondents were asked to rate the quality or influence of articles on applied sports biomechanics in these journals on the 5 point anchored scale: 4-Likely Superior Quality or

Impact, 3-Likely High Quality or Impact, 2-Likely Moderate Quality or Impact,1-Likely Low Quality or Impact, or 0 Unknown Quality or Impact.
The survey also asked respondents to report what they thought were the three most influential applied sports biomechanics textbooks ever written, as well as the three most influential scientific articles in applied sports biomechanics. Demographic data collected were age, primary job responsibility, and country of residence. Journal ratings were compiled and correlated with IF for all journals with an IF in 2008 (n = 24). Statistical significance was accepted at the $P < 0.05$ level. Descriptive data on rankings are reported as mean ± SD.

RESULTS: Fifty-two responses were received from 28 different countries. The mean age was 41 ± 11 years with primary job descriptions of applied researcher (52%) and teacher (31%). Few respondents reported their job as primarily basic research (7%) and other (10%).
Descriptive data on the influence ratings and IF for the journals are listed in Table 1. There was a weak (r_{22} =0.48) correlation between mean ratings and the IF. Forty-seven different books were nominated as "most influential" textbooks in five different languages, and ninety-two different articles were nominated. Table 2 lists the top four nominated publications.

Table 1. Journal prestige ratings (Mean ± SD) in Applied Sports Biomechanics and 2008 Impact Factor (IF)

Journal	Rating	IF	Journal	Rating	IF
J Appl Biomech	3.3 ± 0.9	1.2	*Res Q Exerc Sport*	1.9 ± 1.3	1.2
J Biomech	3.3 ± 0.9	2.8	*Sports Eng*	1.9 ± 1.4	----
Sports Biomech	3.1 ± 1.0	0.5	*Clin J Sports Med*	1.8 ± 1.3	1.6
J Sports Sciences	2.8 ± 1.3	1.7	*J Atl Training*	1.7 ± 1.3	1.7
Med Sci Sports Ex	2.8 ± 1.4	3.4	*J Sports Med Ph Fit*	1.7 ± 1.2	0.7
Br J Sports Med	2.6 ± 1.0	2.1	*J Oth Sports Ph Ther*	1.6 ± 1.3	1.9
Proc: ISBS Conf	2.5 ± 1.2	----	*Perc Mot Skills*	1.5 ± 1.2	0.4
Am J Sports Med	2.5 ± 1.4	3.6	*J Sport Rehab*	1.4 ± 1.2	0.4
Clin Biomech	2.4 ± 1.2	2.0	*J Ex Sci Fit*	1.3 ± 1.2	----
J St Cond Res	2.3 ± 1.2	0.8	*Int J Sp Sci Coach*	1.3 ± 1.3	----
J EMG Kine	2.3 ± 1.3	1.8	*Int J Ap Sports Sci*	1.3 ± 1.4	----
Eur J Ap Physio	2.2 ± 1.4	1.9	*Jap J B Sports Ex*	1.0 ± 1.3	----
J Sci Med Sport	2.1 ± 1.3	1.9	*Res Sports Med*	1.0 ± 1.1	----
Int J Sports Med	2.1 ± 1.4	1.6	*Int J Sp Hlth Sci*	0.9 ± 1.1	----
J Sports Sci Med	2.0 ± 1.2	0.6	*Biology Sport*	0.7 ± 0.9	0.1
J Hum Mov Stud	2.0 ± 1.2	----	*Ap Res Co Athl An*	0.7 ± 1.1	----
Eur J Sport Sci	2.0 ± 1.2	0.8	*Kor J Sports Biomec*	0.6 ± 1.1	----
Sc J Med Sci Sports	1.9 ± 1.4	2.3			

---- Indicates no IF for 2008.

Table 2. Top Four Books and Journals Publishing Articles Nominated as Most Influential in Applied Sports Biomechanics

Books	Percentage of Nominations
Hay 1993 *Biomechanics of Sports Techniques*. Prentice Hall.	22
Winter 2009 *Biomechanics and Motor Control of Human Movement.* Wiley.	14
Kreigbaum & Bartels 1995 *Biomechanics: A Qualitative Approach. . .* Benjamin-Cummings	5
Knudson & Morrison 2002 *Qualitative Analysis of Human Movement.* Human Kinetics.	5
Articles	**Percentage of Nominations**
Journal of Biomechanics	20
Medicine and Science in Sports and Exercise	10
Journal of Sports Sciences	9
Journal of Applied Biomechanics/International Journal of Sport Biomechanics	9

Note: The most recent edition of nominated books is reported.

DISCUSSION: This study confirmed previous observations of a weak (r^2 = 12-23%) association (Knudson & Chow, 2008; Knudson & Ostarello, 2008) between the IF and scholar ratings of influential journals in applied sports biomechanics. This was also consistent with other studies that report weak or no correlations between the IF and disciplinary ratings of journals (Bensman, 1996; Donohue & Fox, 2000; Sellers et al. 2004). Journal ratings in this study and others (Knudson, 2007; Knudson & Ostarello, 2008) are more appropriate estimators of journal prestige and influence in applied sports biomechanics than the IF. The results of the open-ended nominations of influential articles supported the journals most highly rated. Articles nominated tended to be published in the journals ranked in the top journals in Table 1.

A surprising finding was the large variability or limited consensus on what were influential books and articles. There were 47 different books published between 1969 and 2009 that were nominated as the “most influential” books ever written. Influential articles were published in 35 different journals or proceedings between 1899 and 2009. The diversity of responses was likely related to the specific sport interests of the respondents. This was similar to previous studies that reported respondents ratings were strongly influenced by their interest area within biomechanics (Knudson & Chow, 2008). Seven articles received two nominations: three experimental papers (Cavagna et al. 1968; Feltner & Dapena, 1986; Fleisig et al., 1995) and four review papers (Bartlett, 1997; Novacheck, 1998; Putnam, 1991, 1993). Eighteen respondents declined to nominate outstanding articles because they thought it was too difficult or not possible. Some respondents nominated theoretical articles related to their movement interests rather than a paper with explicit application to sports biomechanics. Given this variability, it appears that, as a group, respondents who specialize in sports biomechanics do not share a clear understanding of “applied” sports biomechanics.

The very small response rate in the present study means the sample cannot be considered as representative of the ISBS membership or typical sports biomechanics scholars. Despite these limitations scholars can use these ratings and other publishing ratings to select outlets for publication in applied sports biomechanics research. There was little agreement among the respondents on influential research in applied sports biomechanics. This may represent an opportunity for ISBS to make recommendations or standards on the factors desirable for biomechanical research to be truly applicable to sports.

CONCLUSION: The data support the conclusion that the IF is a poor index of the prestige or influence of journals in the area of applied sports biomechanics. Applied researchers affiliated with ISBS who were inclined to respond to this survey had diverse views about influential books and seminal articles in sports biomechanics. There was also variability in journal ratings and nominations of influential literature possibly because the respondents had specific sports biomechanics research interests.

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KINEMATIC AND KINETIC PATTERNS IN OLYMPIC WEIGHTLIFTING

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The purpose of this study was to identify lower extremity kinematic and kinetic patterns during weightlifting movements and to compare them across different external loads. Subjects completed multiple sets of the clean exercise at various percentage loads. Principal component analysis (PCA) was used to extract kinematic and kinetics patterns of the hip, knee, and ankle joint across the loads. These patterns were then compared across joint and percentage load. Results indicate that lower extremity kinematics and kinetics can be characterized through combinations of PCA-derived patterns. Patterns differed predominantly between joints, but not across percentage loads. The results point to joint-specific lower extremity function during Olympic weightlifting and quantified important technical aspects.

KEYWORDS: principal component analysis, joint coordination, movement patterns

INTRODUCTION: Knowledge of task-inherent biomechanics, such as joint kinematics or kinetics, provides important information for technique training in sports. Success in Olympic weightlifting depends in large part on optimal control and coordination between the joints of the lower extremity (Baumann et al., 1988, Enoka 1988, Hakkinen et al., 1984). Relatively few studies, however, have examined lower extremity joint function during weightlifting movements (Baumann et al., 1988, Enoka 1988, Hakkinen et al., 1984). Moreover, these studies have largely relied on the analysis of discrete peak biomechanical variables. While these variables can provide information about general magnitudes of motions and moments etc. they do not account for the complex interaction between the multiple degrees of freedom that need to be controlled to successfully lift maximal weights during weightlifting movements. Principal component analysis (PCA) is a method that can quantify common synergistic joint coordination patterns across a variety of movements and thus addresses this problem.

In addition to the dearth of information about coordinative patterns during weightlifting movements, surprisingly little is known about load-dependent changes in lower extremity mechanics (Enoka, 1988). Yet knowledge of how these patterns change across loads at each of the lower extremity joints would facilitate a better mechanistic and technical understanding of weightlifting movements. The purpose of this study was thus two-fold; 1) to identify lower extremity kinematic and kinetic patterns during weightlifting exercise and 2) to compare these patterns across joint and load. To this end we used PCA to extract principal patterns of the lower extremity joints during the pull-phase of the clean and compared the extent to which these patterns differed between the hip, knee, and ankle joint across a variety of loads.

METHOD: Ten subjects participated in this study. All subjects participated in a training program that involved weightlifting exercises and were deemed technically competent and representative of collegiate-level lifters by a national USA Weightlifting coach. All subjects provided written informed consent approved by the University's IRB.

Subjects completed a brief warm-up that included lifting light loads up to 50% of their self-reported one repetition maximum (1-RM) for the clean exercise. After the warm-up, subjects performed 2-3 repetitions at 65%, 75%, and 85% of 1-RM with approximately 2-3 minutes rest between each set. Kinematic and kinetic data were collected during each set. Kinematic data were acquired from reflective markers attached to the subjects body with a 6-camera

Vicon motion capture system that sampled at 250 Hz. Kinetic data were collected at 1,250 Hz from two Kistler force plates that were built into an 8'x8' weightlifting platform. Kinematic and kinetic data were filtered at 6 and 25 Hz, respectively. Euler angle rotation sequences were used to calculate ankle, knee, and hip joint angles. Kinematic and kinetic data were combined with anthropometric data and used to solve for net internal ankle, knee, and hip joint moments with an inverse dynamics approach. Moments were normalized to body height and weight. Data were calculated for right leg sagittal-plane variables and time-normalized to 100% of the lift phase (i.e. from the time the barbell left the platform to the time the vertical ground reaction force fell below 10 Newton's at the end of the second clean pull-phase).
For each of the three joint rotations and three joint moments, the time-normalized waveforms for the three sets clean trials of each individual were subjected to a PCA. The input to the PCA for the kinematic and kinetic analysis thus comprised the time-normalized waveforms for all subjects, joints, and lift conditions (i.e. 10 subjects x 3 joints x 3 lift conditions = 90 waveforms), with the values at each 1% time-normalized increment considered the "variables" in the PCA. This yielded a 90 waveforms x 100 "variables" matrix for the joint rotations and moments. From these waveform matrices, principal patterns were extracted using a covariance matrix decomposition method. Only principal patterns that explained nontrivial proportions of the waveforms were retained for analysis. The retained patterns were each normalized to unit vectors and projected onto each original waveform. The sum of these projections over the entire lift phase gave a set of principal pattern scores that expressed the extent to which each pattern was present in the individual waveforms for each subject, joint, and condition. These scores were then used for statistical analysis.
Separate 3 (joint) x 3 (condition) repeated measure ANOVAs were used to test for differences in principal pattern scores. Huynh-Feldt adjustments were made when assumptions of sphericity were not met. The α-level for statistical significance was set at 0.05. In the absence of significant interactions, data were pooled across joint and/or conditions for post hoc testing and compared with bonferroni-adjusted paired t-tests.

RESULTS: Main effects for principal kinematic pattern scores were observed for PP1 and PP2 (Figure 1a), which captured a general extension and an extension-flexion-extension motion, respectively (Figure 1b). More specifically, PP1 scores for the hip and knee were greater than for the ankle, whereas PP2 scores differed between all joints and were greatest for the knee, intermediate for the ankle, and smallest for the hip (Table 1).

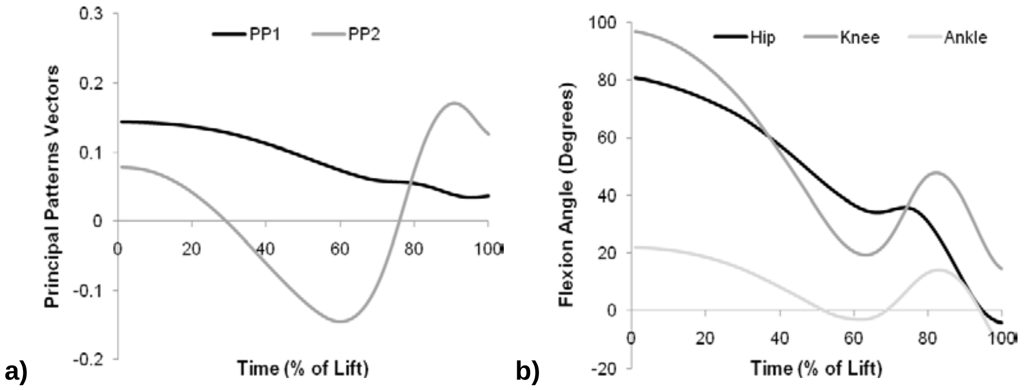

Figure 1. a) Kinematic principal patterns normalized to unit-vectors; b) joint angles during the pull-phase of the clean at 85% of 1-RM (Note: positive angles indicate joint flexion).

Table 1. Principal pattern scores across joint and load

		PC Scores				
		Kinematic		Kinetic		
Joint	Load	PP1	PP2	PP1	PP2	PP4
Hip	65	534.9±97.2*	-16.5±31.8*†	87.3±19.7*†	18.0±16.7	17.1±5.7
	75	530.0±115.0*	-38.7±29.0*†	94.5±22.5*†	20.7±16.2	17.5±3.9
	85	525.3±122.8*	-25.7±36.0*†	92.9±21.0*†	24.5±21.9	18.9±3.9
Knee	65	570.5±153.8*	111.4±25.6*‡	2.6±11.9*‡	20.7±13.2	22.6±11.3*
	75	570.0±142.4*	105.7±29.6*‡	6.6±16.1*‡	23.9±17.2	21.9±10.0*
	85	572.9±129.1*	114.6±32.1*‡	5.6±10.5*‡	19.2±14.9*	25.1±9.7*
Ankle	65	98.8±38.6†‡	41.8±17.5†‡	28.9±9.0†‡	34.9±6.9	10.9±10.8†
	75	112.0±47.5†‡	40.5±16.9†‡	30.8±10.0†‡	36.3±7.4	11.1±8.5†
	85	106.0±38.3†‡	43.6±17.3†‡	34.6±11.6†‡	46.1±8.4†	12.8±7.5†

* p<.05 vs. Ankle, † p<.05 vs. Knee, ‡ p<.05 vs. Hip

Main effects for principal kinetic pattern scores were observed for PP1 and PP4 (Figure 2a), which captured a general extension moment and an extension-flexion-extension moment transition, respectively (Figure 2b). More specifically, PP1 scores differed between all joints and were greatest for the hip, intermediate for the ankle, and smallest for the knee (Table 1). PP4 scores differed only between the knee and the ankle in that the PC scores was greater for the knee. Further, an interaction indicated that the PP2 score (Figure 2a), which captured an extension moment peak during the final part of the movement (Figure 2b), was greater for the ankle than the knee during the 85% condition (Table 1).

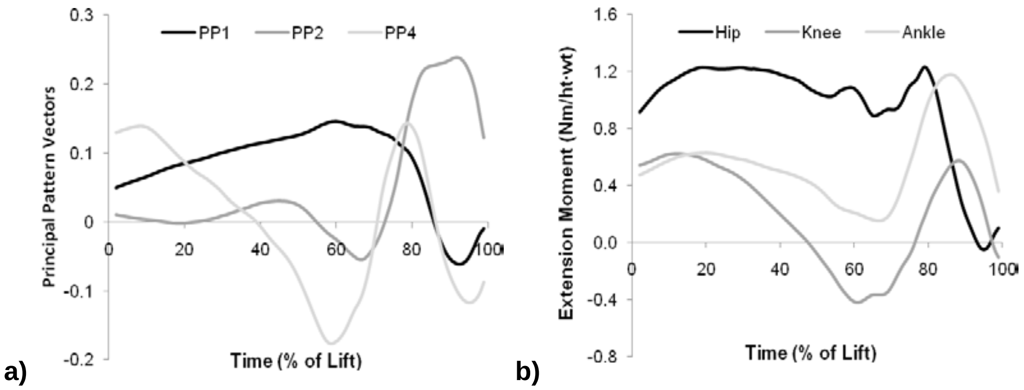

Figure 2. a) Kinetic principal patterns normalized to unit-vectors; b) net internal joint moments during the pull-phase of the clean at 85% of 1-RM (Note: positive moment indicates a net internal joint extension moment).

DISCUSSION: The purpose of this study was two-fold; 1) to identify lower extremity kinematic and kinetic patterns during weightlifting exercise and 2) to compare these patterns across joint and load.
The extracted kinematic patterns captured a general extension motion and an extension-flexion-extension transition. The general extension motion pattern was more prominent for the hip and knee than for the ankle. This pattern seems to reflect the fact that during the pull-phase of the clean the hip and knee joint move through a larger range of motion (Baumann et al., 1988). The second kinematic pattern exhibited a distinct hierarchy between joints and was largest for the knee, intermediate for the ankle, and smallest for the hip. The extension-flexion-extension characteristic of this pattern seems to reflect the double-knee bend transition between the first and second pull of the clean (Baumann et al., 1988, Enoka 1988, Hakkinen et al., 1984), which would explain why this pattern is most prominent at the knee.

The extracted kinetic patterns captured a general extension moment, an extension-flexion-extension moment transition, and an extension moment peak during the final part of the movement. The general extension moment displayed a distinct hierarchy in magnitude between joints and was largest for the hip, intermediate for the ankle, and smallest for the knee. The magnitude of the hip moment correlates well with the magnitude of the weight lifted during weightlifting competition (Baumann et al., 1988) and would indicate that this is a very important characteristic. Similar to the kinematic analysis, the kinetic PCA also extracted an extension-flexion-extension pattern that appeared to reflect the double-knee bend transition. Since this kinetic pattern was greater for the knee than the ankle, the results would indicate that neuromuscular control of the knee joint is more important during this phase than that of the ankle. The analysis also captured an extension moment peak during the final part of the movement that was greater for the ankle than the knee during the 85% of 1-RM load condition. This interaction indicates that the contribution of the ankle extensor musculature during the final pull phase in weightlifting is more prominent at higher load percentages.
Collectively, the PCA-derived patterns are able to describe lower extremity function during weightlifting exercise. Hip function during weightlifting is characterized by a general extension motion and large extension moment, which in combination indicates a large requirement of mechanical work from the hip extensor muscles. Knee function was most notably characterized by an extension-flexion-extension pattern in both joint motion and joint moment, which underscores the technical importance of the double knee bend during weightlifting. Lastly, ankle function consisted of small amount of angular excursion, intermediate extension-flexion-extension motion and moment magnitudes. Interestingly, when compared to the knee joint, the magnitude of ankle joint moment was greatest at higher loads, which underscores the importance of ankle function as lift weight increases.

CONCLUSION: The results indicate that lower extremity kinematics and kinetics can be described by PCA-derived patterns. Kinematic patterns differed between joints, but appeared robust and invariant in response to changes in external load. Although two kinetic patterns differed between joints only, one kinetic pattern exhibited more complex behaviour in that it differed across joint and load. Collectively, these patterns were able to provide technical perspectives on lower extremity function during weightlifting exercise.

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LUNGE FORCES AND TECHNIQUE OF JUNIOR SQUASH PLAYERS

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KEYWORDS: kinetics, squash, lunge performance, ground reaction force, biomechanics.

INTRODUCTION: The lunge movement is used regularly in squash, as well as in other sports such as badminton and fencing, and the ability to complete a controlled lunge quickly can be a crucial part of the game (Cronin, McNair, & Marshall, 2003). The lunge has been recognised as placing high physical demands on the lower limbs, with vertical ground reaction forces (GRF) exceeding 2.5 times body weight during a badminton lunge (Kuntze, Mansfield, & Sellers, 2010). There have been a number of studies examining the forces produced by adult athletes performing a lunge movement (Lees & Hurley, 1994; Kuntze, Mansfield, & Sellers, 2010), however, to date there is very little information on the kinetics or kinematics of the squash lunge technique as performed by junior athletes. The aim of this study was to quantify and compare the ground reaction forces produced by junior squash players while performing a simulated forehand and backhand lunge shot.

METHOD: Nine male junior squash players participated in this study (age 14.53 ± 1.96 yrs, height 1.62 ± 0.05 m, body mass 54.2 ± 9.3 kg). Subjects were divided into 2 groups according to their playing ability: Experienced Juniors (EJ), n=4 and Developing Juniors (DJ), n=5. Subjects were required to run a distance of approximately 4 m before lunging forward onto one leg, performing a ghost shot, then pushing off and returning to the start position. Three forehand (FH) and three backhand (BH) trials were performed by each subject. Data for each trial were collected via 12 Vicon MX-13 cameras (OMG, England) and Kistler force plates (Kistler, Switzerland) through Vicon Nexus software at 500 Hz. Each subject was marked up according to the standard lower-body Vicon Plug-in Gait model. The 3rd trial for each subject was chosen for analysis to eliminate any learning effect. The variables chosen for analysis were: knee angle at touchdown and takeoff; max knee flexion; approach angle; total contact time; max horizontal and vertical GRF; and vertical impulse. The vertical GRF component was further broken down into 6 variables, as classified by Lees & Hurley (1994) (identifiable in Figure 1); A: initial impact peek (heel strike), B: impact loading, C: amortization (force reduction), D: loading (weight acceptance), E: drive-off, PFR: peak force reduction, difference between B and C. Statistical significance was assessed using a two-way ANOVA (SPSS) with 2 factors corresponding to ability (EJ and DJ) and stroke (FH and BH). Significance was set at p < 0.05. Post-hoc analysis was done via a student t-test.

RESULTS: A typical vertical (Fz) GRF curve for a subject from the EJ and DJ group during a FH lunge can be seen in Figure 1.

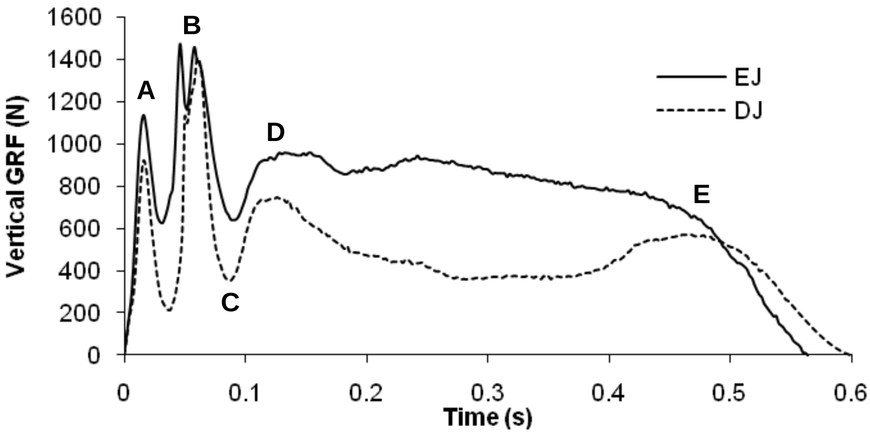

Figure 1. A typical Fz GRF curve for a subject from the EJ and DJ group during a FH lunge.

There were no significant differences found between the forehand and backhand strokes for any of the variables evaluated. A summary of the mean data (relative to body weight) for the 6 vertical GRF variables are presented in Table 1. Initial impact peak (A) was significantly higher, while impact loading (B) and peak force reduction (B-C) was significantly lower for the EJ group compared to the DJ group (see Table 1). There were no other significant differences between abilities for any of the other kinetic or kinematic variables measured.

Table 1. Mean vertical (Fz) forces relative to body weight (N) during the lunges.

		Experienced Junior	Developing Junior	*p* value
A	Initial impact peak	2.49 ± 0.48	1.92 ± 0.26	0.017*
B	Impact loading	2.44 ± 0.30	3.22 ± 0.38	0.006*
C	Amortization	1.01 ± 0.16	0.94 ± 0.16	0.473
D	Loading	1.57 ± 0.18	1.66 ± 0.09	0.271
E	Drive-off	1.34 ± 0.09	1.26 ± 0.09	0.157
B-C	Peak force reduction	1.43 ± 0.26	2.28 ± 0.37	0.003*

*$p < 0.05$

DISCUSSION: There were no significant differences between any variables when comparing FH to BH, suggesting that even though subjects were using alternate legs, they produced similar force curves/magnitudes and performed the lunge with similar kinematics. The lower initial impact forces (A) and higher impact loading forces (B) in the DJ group is most likely the result of landing with a slightly straighter leg, combined with a more flat-footed strike than the EJ group. The DJ group tended to begin knee flexion later, after the foot was completely flat on the ground, suggesting the inexperienced players had not yet developed the appropriate coordination and movement skills, or strength, to reduce this aspect of the impact force (Lees & Hurley, 1994). Furthermore, the greater peak force reduction seen in the DJ group may also indicate an inferior ability of the developing players to efficiently utilize the high impact loading experienced during the lunges.

CONCLUSION: This study showed that developing junior squash players tended not to be able to efficiently utilise the high impact forces generated during impact loading. They appear to have not developed the necessary coordination and movement skills to reduce the affect of impact loading. It was also shown that there were no significant differences in the measured variables between forehand and backhand lunges as performed by this group of male junior squash players. The main limitation of this study was the small sample size of each group. Further investigation with a larger sample size is recommended in order to gain a greater understanding of the lunge forces and technique of junior squash players.

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WITHIN SUBJECT VARIABILITY ANALYSIS REVEALS A TRANSITION POINT FOR THE LONGSWING ACROSS AGE GROUPS

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This research aimed to observe changes in the within subject variability of the longswing performance and coordination across age groups in gymnasts divided by their competition level, from younger (group 1) to experts (group 5). Data were collected by two video cameras. Performance and coordinative within subject variability were calculated by the standard deviation (inter-trial variability) and the deviation phase (intra-trial variability). Results only showed significant group differences for within subject variability (inter- and intra-trial) in the SD P3H-P3S. In addition, group 4 (14.78±0.57 yrs) showed both large inter-trial variability in the upswing shoulder flexion (P3) and large intra-trial variability during hip and shoulder extension (P2) and P3. Such large variability in group 4 suggests a transition point towards the experts´ performance and coordination (19.96±3.37 yrs).

KEYWORDS: motor learning, performance, coordination, gymnastics.

INTRODUCTION: An effective motor strategy (performance and coordination) should to be discovered during the process of skill acquisition. Coordination has been defined as the stable spatial-temporal relationship among limb segments or joints to achieve the task's goal (Irwin & Kerwin, 2007). Discovering new modes of coordination may involve undergoing a transition from one stable form of coordination to another (Handford et al., 1997). High within subject variability is characteristic of a system in transition (Clark & Phillips, 1993). It was suggested that within subject variability conforms to a U-shaped graph as a function of skill progression (Wilson et al., 2008), because high within subject variability were observed in beginners as well as experts. Sport skills represent an ideal situation to assess changes in performance, coordination, and within subject variability at different levels of expertise. The 'regular' longswing on high bar in gymnastics was selected as the focus of our research because this skill is identified as a fundamental basic skill (Irwin & Kerwin, 2007). The principal aim of this study was to assess changes in the within subject variability of the longswing performance and coordination across different competition age groups. Additionally, the U-shaped fit of the within subject variability across the groups was analyzed.

METHOD: Five competition age groups: group 1 (8.92±0.85 years); group 2 (11.08±0.67 years); group 3 (12.88±0.50 years); group 4 (14.78±0.57 years), and group 5 (19.96±3.37 years) were used to classify participants (113 male gymnasts). All participants gave informed consent. The study was approved by the local ethics committee.
We defined 3 events independently for hip (H) and shoulder (S) angle joints (Figure 1): the smallest angle during downswing (P1H, P1S); the largest angle after P1 (P2H, P2S); and the smaller angle during upswing (P3H, P3S). We focused our study in P2 and P3 given that functional phases of the longswing are defined by these events (Arampatzis & Brüggemann, 1999). The total maximum elevation of the center of mass on the downswing, initial position (Pi), and the maximum elevation on the upswing, final position (Pf), were used to infer the total path of the swing and defined as swing amplitude (Figure 1). Participants were asked to perform ten consecutive swings from a quiet starting swing position under the bar. They were filmed with two digital video cameras located on the participant's right side and in front of them describing a 90 degrees angle between their optical axes. The best three consecutive swings, selected qualitatively by an expert coach, were analyzed for each participant. The videotaped images captured at 50 Hz were manually digitized (Kwon3D, Young-Hoo Kwon & Visol, Inc). Flexion-extension angular displacement and velocities for the hip and shoulder in the sagittal plane were computed. Body position angle was defined as the angle formed by

the line connecting the center of mass with the middle of the grasping hand and the vertical (z-axis) (Arampatzis & Bruggemann, 1999) (Figure 1). We calculated the location of the center of mass differently depending on the gymnast age (Jensen & Nassas, 1988; Jensen, 1989; De Leva, 1996).

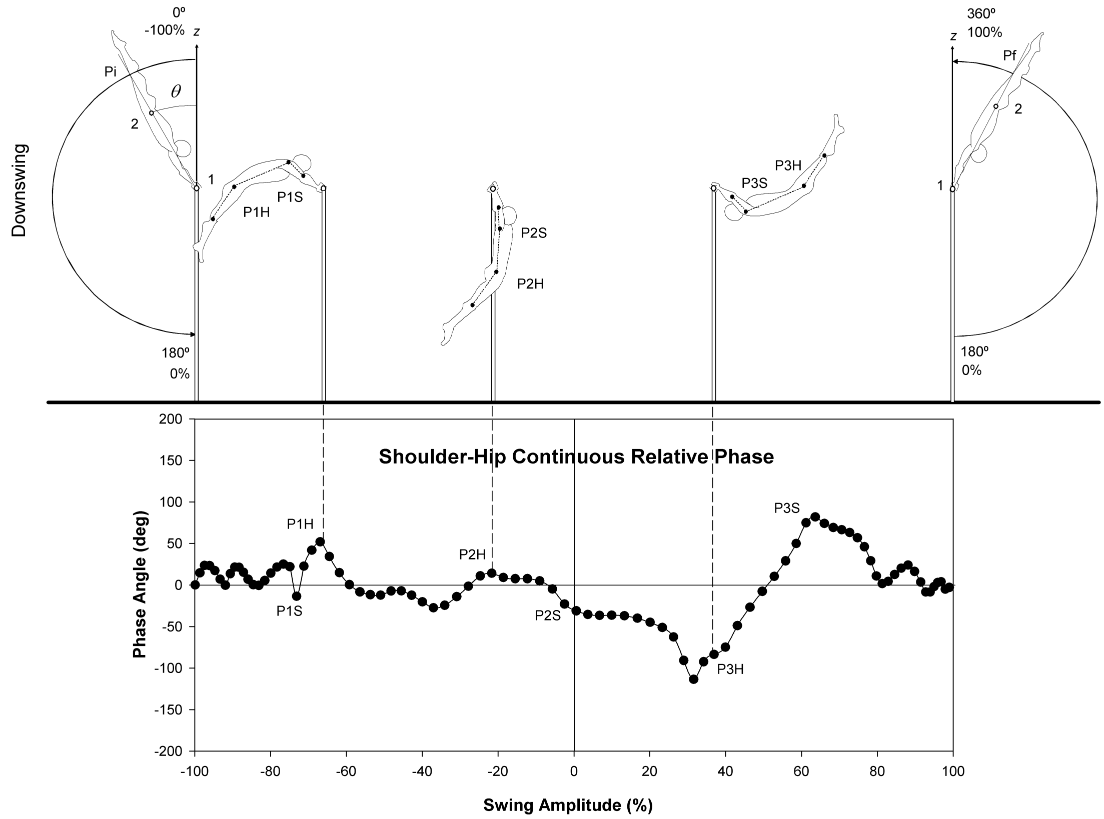

Figure 1. In the upper section, swing amplitude defined in reference of the body position angle (θ) delimited by z axis, middle grasping hand landmark (1) and the center of mass (2). Additionally, we illustrated the initial position (Pi), final position (Pf) and swing events (P1, P2, and P3) from the hip (H) and shoulder (S) joints. In the lower section, the continuous relative phase between the hip and shoulder joints during a longswing of an expert gymnast (group 5) is represented. For simplicity, H and S events have been represented at the same instant of time for P1-P3 in the upper section.

Performance was described by swing amplitude and events, while coordination was assessed using the inter-joint reversal points (P1H-P1S, P2H-P2S, and P3H-P3S) and the absolute difference in the continuous relative phase between contiguous events of the two joints. Each continuous relative phase was obtained by subtracting the phase angle of the distal joint (hip) from that of the proximal (shoulder) (Clark & Phillips, 1993; Hamill et al., 1999). In turn, the phase angle (φ_i) was calculated from the normalized angular displacement (θ) and angular velocity (ω) using $\varphi=\tan^{-1}(\omega / \theta)$. In Figure 1 (lower section), the continuous relative phase of an expert gymnast was depicted for a longswing in high bar. Relative phase around -20% was close to 0º indicating that the hip and shoulder moved in synchrony or they did not move remaining with the same angle values. This in-phase relationship became clearly out-of-phase led by hip flexion around 30%. Around this point in the upswing, the hip achieved the maximum flexion (P3H). Subsequently, the coordination changed faster to a positive out-of phase mode led by the shoulder's flexion while the hip initiated slower its extension. Performance and coordination within subject variabilities were assessed in two different ways: inter-trial variability (standard deviation between the trials of the participants, SD) and intra-trial variability (coordination changes within the trial described by the deviation phase, DP).

One-way ANOVAs were used to examine differences across the competition age groups in standard deviation and deviation phase. Tukey post-hoc comparisons between groups were conducted when appropriate. Statistical significance was set at $p<.050$ level; however, p values between .050 and .100 were also discussed. When normal distribution (Kolgomorov-Smirnov test) and homogeneity of variance (Levene test) were verified, parametric statistics were used; else non-parametric test were used. All tests were performed with Systat 11.0 and SigmaStat 3.1 (Systat Software, Inc., San José, CA, USA).

RESULTS: Results of this study found group differences in the SD P3H-P3S (H=9.47, p=.050). Despite pair comparisons not yielding significance, it's important to notice that group 2 and group 3 had more consistency than the other groups. The intra-trial variability variables (DP P2H-P2S and DP P3HP3S) did not show significant differences between the competition age groups; however, both variables showed a tendency to differ between groups (DP P2HP2S: H=9.26, p=.055; and DP P3H-P3S: $F_{4,334}=2.15$, p=.074). Compared to group 1-group3, deviation phase in P2 was larger for group 4 and smaller for group 5 (Figure 2e), while deviation phase in P3 was higher for the group 4 and group 5 (Figure 2j). As seen in Figure 2, group means of the inter-trial variability in P2 had a better U-shape fit than in P3 variables. Interestingly, group 4 showed the largest within subject variability in all P3 variables except SD P3H compared to the rest of the groups.

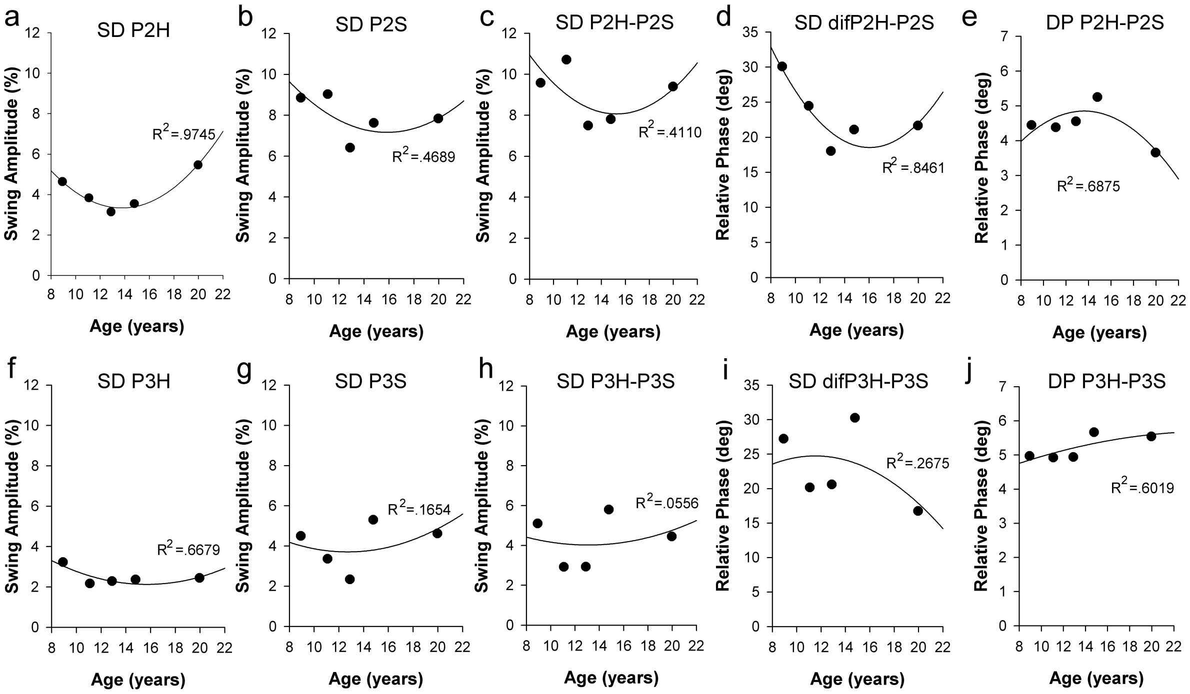

Figure 2. Graphs illustrate group means of the inter-trial variability of performance (a, b, f, and g) and coordination (c, d, h, and i), and intra-trial variability of coordination (e and j). P2 variables are depicted in the upper section and P3 variables in the lower section. Quadratic curve fit for data points and R^2 are also included.

DISCUSSION: Two distinct patterns were observed: a U-shaped fit, and a large deviation for group 4. A better U-shaped fit was found in graphs obtained from P2 inter-trial variability variables supporting the results presented by Wilson et al. (2008). In contrast, large group 4 deviations were found in graphs of the inter-trial variability for the P3 variables and the intra-trial variability variables. It could be suggested that group 4 represents a transition point in the process to achieve the expert coordination mode of group 5. In agreement with these results, three of these variables (SD P3H-P3S, DP P2H-P2S, and DP P3H-P3S) also showed a tendency for group 4 to have increased variability suggesting again a transition point. These results could be interpreted on the basis of previous research (Teulier et al., 2006) proposing that when the initial behavior does no longer improve the task, the performer needs to change motor strategies. However, the swing amplitude between group 4 and group 5 were not statistically different. Therefore an alternative explanation is provided: the transition phase in group 4 could be due to increased demands of the sport. Sport demands at the age range of the group 4 drastically change by including flight elements and dismounts learning.

CONCLUSION: Large increases in within subject variability in P3 and intra-trial variability for the group 4 indicated a transition point towards the expert coordination mode of group 5. Two different arguments are proposed to explain this transition point occurrence: (1) motor strategies adopted until group 4 did not improve anymore the skill level (Teulier et al., 2006); and (2) the increased demands of the sport incorporate the longswing into more complex tasks.

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CHARACTERISTICS OF JOINT MECHANICAL WORK IN MALE AND FEMALE ELDERLY DURING WALKING IN CONSIDERATION OF VELOCITY

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KEYWORDS: aging, walk, kinetics, ANCOVA

INTRODUCTION: Previous studies about how the elderly walk have reported their kinematical and kinetic characteristics by comparing their motions with those of young walkers (Murray et al., 1969; Hageman and Blanke, 1986; Blanke and Hageman, 1989; Winter et al., 1990; Kaneko et al., 1991; Judge et al., 1996). The results from these studies contain both the effects of aging and walking velocity because walking velocities of the elderly and the young differed from each other and the velocity strongly affects most biomechanical variables. In addition, the change in walking motion with aging after sixty has rarely been reported, and the differences between elderly male and female walkers were also not clear. This study seeks to clarify the differences in the characteristics of joint mechanical output and contribution during walking between different age groups and sexes for the elderly in consideration of walking velocity.

METHOD: The subjects were 213 healthy Japanese male and female elderly who were divided into six groups according to their age and sex (Table 1). They were instructed to walk about 10m at four self-selected speeds (Slow Walk (SW), Normal Walk (NW), Fast Walk (FW), and Maximum-speed Walk (MW)). We videotaped them during walking with a digital VTR camera at 60fps in order to analyze their motion in the sagittal plane. The ground reaction forces on the right foot were measured by a force platform installed below the walkway. Two-dimensional coordinates of the eight body landmarks were obtained by using a video digitizing system (Frame-DIAS, DKH Co., Ltd., Japan). Four trials (one for each walking speed) per each subject were analyzed. The coordinates were smoothed by a fourth-order, zero-phase-shift Butterworth digital filter at the optimal cut-off frequencies that were derived from residual analysis (Winter, 1990). After synchronizing the smoothed coordinate data and ground reaction forces, we calculated joint torques at the ankle, knee and hip using a link-segment model based on the inverse dynamics method. Joint torque powers were next calculated by multiplying the joint torque by the joint angular velocity. Finally, joint mechanical work was calculated by integrating the joint torque power over time and the joint contribution to the total lower limb work was expressed as the percentage of each joint work to the total. In order to test the differences in the work and the contribution between the groups without the effect of walking velocity, two-way analysis of covariance (ANCOVA) was done in which the walking velocity was set as the covariate. When a significant effect of age or sex or their interaction was recognized, a multiple comparison (Scheffe test) was done.

RESULTS AND DISCUSSION: Figures 1 and 2 plot the positive work done by the ankle and the hip during one walking cycle. Both the ankle and hip positive work increased with the

Table 1 Profile of subjects

Sex		Range of Age [yr.]	n	Age [yr.]	Standing Height [cm]	Body Mass [kg]
Male	E1	65-69	38	67.5 ± 1.3	162.0 ± 5.0	63.6 ± 7.0
	E2	70-74	42	72.5 ± 1.5	163.2 ± 5.0	64.0 ± 7.0
	E3	75+	24	77.9 ± 2.6	161.3 ± 6.1	62.4 ± 7.2
Female	E1	65-69	47	67.6 ± 1.3	149.2 ± 4.7	55.7 ± 7.8
	E2	70-74	44	72.5 ± 1.4	149.3 ± 5.7	53.4 ± 6.5
	E3	75+	18	77.0 ± 1.7	148.2 ± 4.8	52.6 ± 6.5

walking velocity. Regardless of sex, the ankle positive work at the same walking velocity decreased with age, and that of E3 was significantly smaller than that of E2 ($p<0.05$). In contrast, the hip positive work at the same walking velocity increased with age, and that of E3 was significantly larger than that of E2 ($p<0.05$) and E1 ($p<0.01$). Although the knee positive work also increased with walking velocity, no significant differences between age groups were seen. The contribution of the ankle and knee joint to the total positive work differed with the change of the walking velocity. The contribution of the ankle decreased and that of the knee increased with the increase in walking velocity. In contrast, the contribution of the hip was almost the same at different walking velocities. The contributions of the ankle and the hip differed between the age groups. The contribution of the ankle for E3 was significantly smaller than that for E2 ($p<0.01$) and E1 ($p<0.05$). In contrast, the contribution of the hip for E3 was significantly larger than that for E2 ($p<0.05$) and E1 ($p<0.01$). These results indicated that the function of plantar flexors during walking deteriorates with aging and the hip joint must then exert more power to compensate for the decline in ankle function, and this tendency was seen in both of male and female walkers.

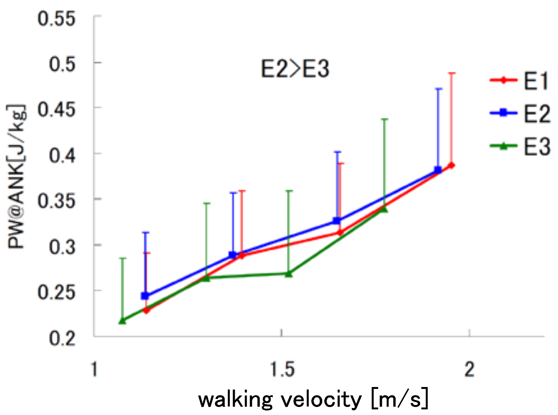

Figure 1 Ankle positive work

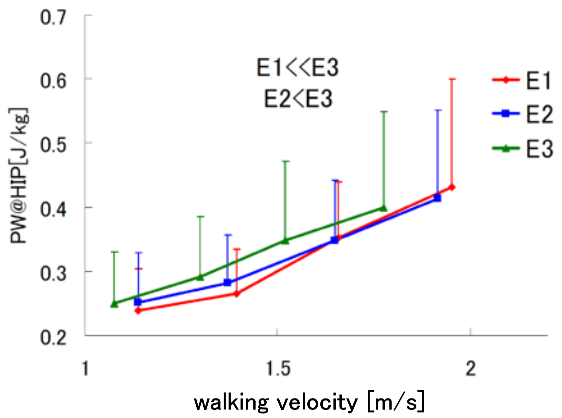

Figure 2 Hip positive work

CONCLUSION: This study examined the effect of age and sex on the characteristics of joint kinetics in consideration of walking velocity by using two-way ANCOVA. The results from the positive work and the joint contribution to the positive work indicated that power generation in the ankle joint deteriorated with aging, and this deterioration was seen in both sexes. This indicates that maintaining the ankle power is very important for male and female elderly to keep their walking ability.

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DIFFERENCES IN RSI AND PEAK GROUND REACTION FORCE FOR DROP REBOUND JUMPS FROM A HANG AND BOX FOR FEMALE SUBJECTS

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The aim of this study was to examine differences between drop jump rebound from a box (DJB) and hanging position (DJH). The volunteers were 14 college aged women who were healthy and physically active. Jumps were assessed from a 30 cm drop onto a force platform. Jump height, contact time, reactive strength index, and peak vertical ground reaction force were compared for the two jumps using a Paired T-test. There were no significant differences in peak ground reaction force or jump height between the DJB and DJH ($p > 0.05$). Contact time was found to be less ($p = 0.033$) and RSI higher ($p = 0.012$) for the DJB. Thus, the DJB would be recommended for power training over the DJH. The results of the current study may aid coaches to prescribe the optimal drop rebound jump position for the training of their athletes.

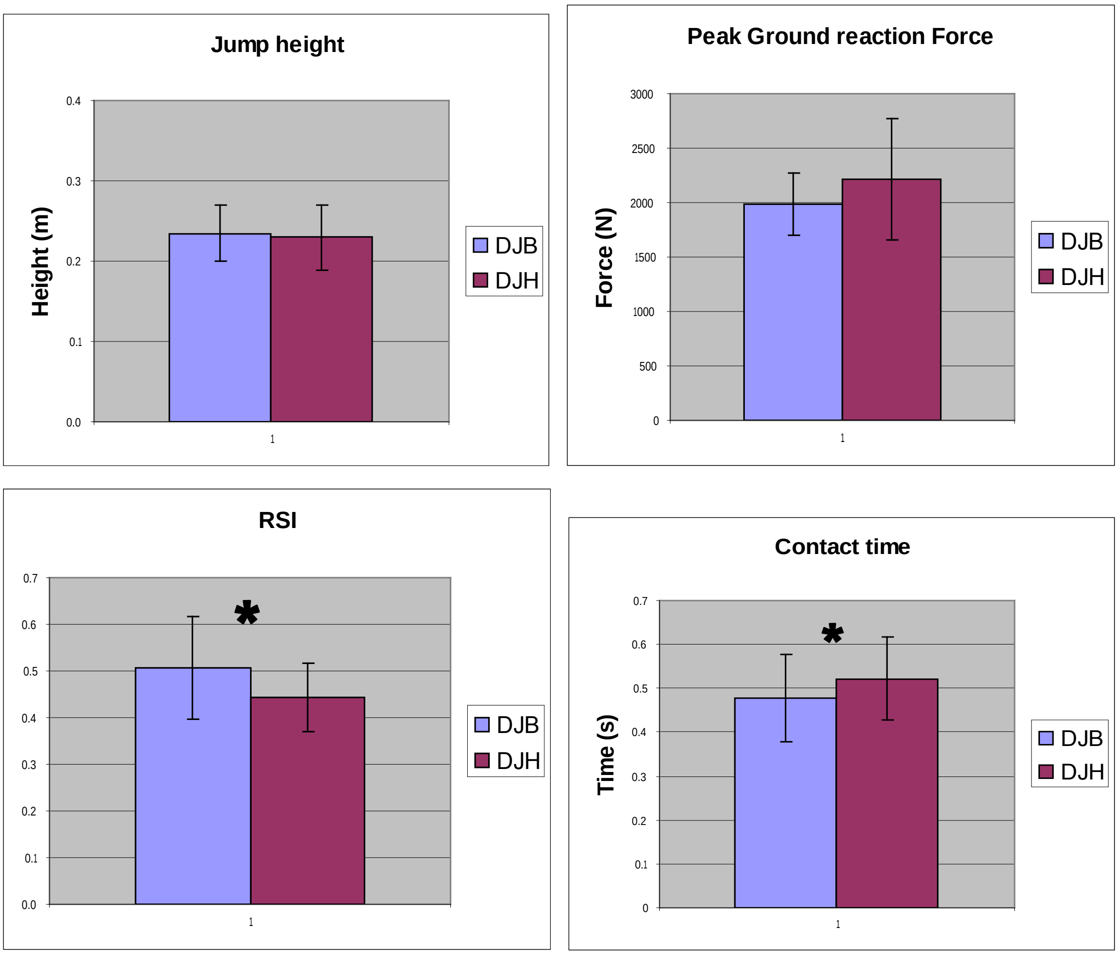
KEYWORDS: plyometrics, stretch shorten cycle, reactive strength index, contact time

INTRODUCTION: Plyometrics is a common training method used to increase power and jumping performance (Ebben, 2010). Rebound jumps initiated with a drop from a box are commonly used for training and assessment of plyometric performance. However, the drop from a box often may involve a hop/shuffle forward off the box, thus altering the actual drop position to the landing surface. A drop that takes place while hanging from a support would eliminate the hop/shuffle and any associated horizontal impact. Difference in jump height and ground reaction forces have been previously investigated during loaded and unloaded drop jumps (Tsarouchas 1994). Thus, this study will investigate the differences between drop jump rebound from a box (DJB) and drop jump rebound from a hang (DJH). Differences in jump height, contact time and thus reactive strength index (RSI) and vertical peak ground reaction forces (VGRF) during take off will be compared between the two jump positions. Anecdotally the hypothesis would be that jump height may be greater for the drop from hang as the efforts may be more easily focused up and not slightly forward.

METHODS: Fourteen female volunteers (Mean ± SD; age = 20.00 ± 3.00 years; height = 1.73 ± 0.89m; body mass = 68.60 ± 10.40kg.) were recruited on a voluntary basis from a variety of sporting backgrounds. All were physically active (trained 3+ days a week) with no lower limb injuries in the previous six months, while experience of plyometric training was varied. Each signed an informed consent form and filled out a readiness for physical activity questioner. Ethic approval was granted by the university Human Subject's Research Review Committee (#HS10-321).
The volunteers abstained from intense training and resistance training for the previous 24 hours. They underwent a warm up on a cycle ergometer for 5 minutes at 60 rpm, which is equivalent to 90 watts. Dynamic stretching and squat jumps were performed followed by four familiarization drop rebound jumps from a 30cm box and four from a 30cm hanging position. A 5 minute rest was taken before measurement of the jumps. Volunteers performed three maximal DJB and DJH both from 30cm in random order. The hands were placed on the head to prevent arm swing after the release from the hanging and box positions. Instructions for the jumps were to jump as fast and as high as possible to ensure the highest RSI (Young et al, 1995).
Jumps were recorded on a force plate (OR6-5-2000, AMTI, Watertown, MA, USA) sampling at 1000Hz. Jump height was calculated using the time in air method (Komi & Bosco, 1978) that assumes the centre of mass was at the same height at take off as at

landing. This assumption seems safe to use as error was calculated to be less than 3% by Frick et al (1991).
Statistical analysis was done using a paired two tailed T-test on Microsoft Excel. The jumps using the highest RSI were considered the best performance and were the ones analysed for all dependent variables. The alpha level selected was $p \leq 0.05$.

RESULTS: T-test revealed no significant difference in VGRF between the DJH (2208 ± 558N) and DJB (1988.7 ± 282.3N) (p=0.18). There was also no significant difference found for height jumped (p=0.36) with mean height for DJH and DJB being 0.229 ± 0.041m and 0.234 ± 0.035m respectively. Significant differences were found for contact time for DJH 0.522 ± 0.095s to 0.477 ± 0.100s for DJB (p=0.03). RSI also differed between DJH and DJB (p=0.01) with means and SDs of 0.44 ± 0.07 and 0.51 ± 0.11 respectively. These are represented in graphic from as seen in Figure1.



*= Statistical significant difference p=0.01 *= Statistical significant difference p=0.03

Figure 1. Means for the jump height, VGRF, RSI, and contact time of the best jump for each subject. (n=14).

DISCUSSION: The major findings of this study reveal no significant difference between DJH and DJB for either height or VGRF. The fact that the jump height is the same for either drop jump position suggests they could be used interchangeably when training for or assessment of jump height. Because training specificity is important, in activities such as blocking in basketball/volleyball where height is most important, training with either type of drop jump would suffice.
With regards to training the stretch shortening cycle both types of rebound jumps were found to be slow stretch shortening cycle jumps as the contact time was >0.25s as noted

by Schmidtbleicher (1992). So unless attempting to improve rate of force development or reducing contact time then the use of either would suffice.

Use of RSI is agreed to be a better measure of power than height jumped alone (Komi, 2000; McClymont, 2007; Schmidtbleicher & Komi, 1992). As the RSI was significantly higher for the DJB than DJH it is a more powerful movement. In addition, Young (1995) suggested that DJB is a valuable performance measure and included it on the list of Strength Qualities Assessment Test battery. The fact that a drop jump causes the athlete to have to absorb their momentum first and then jump makes it a reactive type strength. Training to reduce contact time/increase tendon stiffness, e.g. for sprinters where running velocity is key, may be optimized by using the DJB (Harrison et al., 2004; Comyns et al., 2007).

In the current study, only verbal instruction followed by a demonstration was given, thus a lack of feedback may not have yielded a true maximum performance for the individual jumps. Young et al (1995) gave visual feedback on contact time and height jumped after each jump in addition to verbal instructions and found that the combined feedback resulted in better RSI compared to each feedback strategy separately.

Also the fact that arm swing was not allowed and hands were placed on the head may have lowered the outcome performance scores. Previous studies have shown that arm swing augments jump performance (Harrison & Maroney, 2007; Walsh et al., 2004). Because the current technique (keeping the hands on the head) was not regularly used by the subjects in typical jump performance, it may have affected the outcome as well. Nevertheless, this technique was used to allow the two jumps to be similar in arm use and centre of mass throughout the jump.

CONCLUSION: Both jump techniques yielded similar jump height and VGRF, however the DJB demonstrated significantly higher RSI and lower contact time. However, DJH required more time for set up and unique equipment (elevated hanging bar). In addition, because the DJB exhibited a higher RSI and lower contact times it would be optimal for power training, compared to the DJH. The DJH may ensure that the impact will be straight down and it may add variety to the training. In the field, very few if any sports require absorbing a force straight down, an example might be blocking in volleyball or rebounds in basketball; most other sports will have a horizontal aspect e.g. even a high jump take off has horizontal velocity. In conclusion, DJB seems to be a more practical and a more appropriate drop jump position.

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THE RELATIONSHIPS BETWEEN POSTURAL STABILITY AND FUNCTIONAL ACTIVITY IN OLDER ADULTS

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KEYWORDS: static balance control, functional activity, elderly

INTRODUCTION: Good postural stability is critical for excellent performance, injury prevention, and independent living quality in older adults. The purpose of this study was to investigate the relationships between postural stability and functional activity in the elderly.

METHOD: Thirteen healthy elderly adult males (74.2 ± 5.5 yrs, 159.9 ± 6.6cm, 62.6 ± 8.6 kg) were recruited and screened for lack of any physical impairments or acute musculoskeletal disorders. Postural stability was measured using a portable three-axis force plate (Accusway Plus, AMTI, Watertown, MA, USA), for which the sample rate was set at 100 Hz and the mean velocity, radius and the sway area of the center of pressure (COP) displacement were collected in an eyes-opened condition. Functional activities were evaluated by 6 meter up-and-go and 30 second chair sit-and-up tests. The subject sat on a chair without arm rests (height 46 cm) with their feet shoulder width apart, flat on the floor. The arms were to crossed at the wrists and held close to the chest. From the sitting position, the subject stood completely up, then completely back down; this was repeated for 30 seconds. The total number of complete chair sit-and-up (up and down equaled one stand) was counted. If the subject completed a full stand from the sitting position when the time was elapsed, the final stand was counted in the total. In up-and–go test, a marker was placed 6 meters in front of the chair. The subject started fully seated, hands resting on the knees and feet flat on the ground. On the command, "Go," timing was started, the subject stood and walked (running was disallowed) as quickly and safely as possible to and around the marker (cone), returning to the chair to sit down. Timing stopped as soon as they sat down. The subjects underwent all the tests twice with random order, and the best performance in each activity was recorded. Pearson product moment correlations were calculated to assess the linear relationships among variables and the statistical significances were set at $p < .05$.

RESULTS: Table 1 presents the results of each testing. Table 2 showed that significant correlations were found between up-and-go and the mean velocity, radius and the sway area of center of pressure (COP) displacement, respectively (r = .44~.61, $p < .05$). However, no significant correlation was found between 30 seconds chair sit-and-up and static balance control.

Table 1. Results of balance control and functional activities testing

	COP Radius (cm)	COP Velocity (cm/s)	COP Sway Area (cm^2)	30s Sit & Up (repetitions)	6m Up & Go (minutes)
Mean	0.59	1.28	4.64	12.0	9.0
S.D.	0.14	0.33	2.30	3.0	2.1

Table 2. The correlation among variables

COP displacement	30 seconds chair sit-and-up	6 meter up-and-go
Radius	-.31	.49*
Velocity	-.05	.44*
Sway area	-.37	.61*

* significant correlations between variables.

DISCUSSION: Preventing the elderly from falling and evaluating the falling risk are critical. Measuring the balance control and functional activities of lower extrimities to evaluate the risk of falling are suggested assessments. It is believed that 30 seconds chair sit-and-up and up-and-go tests could be used to measure the abilities of lower extrimities in the elderly (Jones, 2002). This study demonstrated that there were positive correlations between static balance control and up-and-go but not for the 30 second chair sit-and-up. The nature of these functional activities might be the main reason for the differences in correlations with balance control. The chair sit-and-up performance is an indicator of the strength and endurance of legs; however, the 6 meter Up-and-go test may be a better way to measure the postural control ability in the older adults. The cause-effect relationship between postural control and other functional activities still need further study.

CONCLUSION: The functional activity, up-and-go has positive relationship with postural stability indicates that the elderly with better postural stability intend to have better performance in 6 meter up-and-go test.

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THE USE OF UNI-AXIAL GYROSCOPE FOR MONITORING HEEL TILTING VELOCITY DURING SIMULATED ANKLE SUPINATION SPRAIN MOTIONS

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This study investigated the feasibility of using a uni-axial gyroscope to monitor the motion of foot segment. Five male subjects performed supination spraining motion simulated by a mechanical sprain simulator. A uni-axial gyroscope was attached on the shoe surface at the heel position of the right shoe to collect the heel tilting velocity. Optical motion analysis was also used to obtain heel tilting velocity as a standard. The intra class correlation and root mean square error of tilting velocity measured by the two methods are 0.70 – 0.99 and 8.21 – 37.11 deg/s, respectively. The result shows that it is possible to use only one uni-axial gyroscope for monitoring foot segment motion. This monitoring method can be contributed to the currently developing active protection “sprain-free shoe”.

KEYWORDS: motion sensor, ankle kinematic, ankle sprain

INTRODUCTION: Recently, our research group have been working on the development of innovative intelligent sprain free sport shoe for the prevention of ankle sprain injury (Chan, 2006). Before initiating active correction mechanism in case of an ankle sprain, the shoe system measures and monitors the ankle kinematic changes in order to recognize if it is approaching to a sprain injury. Accelerometer, gyroscopes and magnetic sensors can be used to monitor the ankle motion in a real time manner. Despite the small size, lightweight and generally low power of the sensors, it is still a very challenging task to assembly all these on the foot, lower leg and thighs of both limbs for daily use. Therefore, the number of sensors should be minimized and housed into the shoe such that it is suitable for the consumer product. Here we propose an idea of using one uni-axial gyroscope to measure heel titling velocity. Heel tilting is defined as the motion of the foot segment relative to the ground. It is difference from inversion motion, which relatives to the shank. Hence, at least two motion sensors are needed for the measurement of inversion motion. This study is to examine the validity of the gyroscope relative to the optical motion analysis system when using one gyroscope at heel cup to monitor the tilting angle of the foot segment. This can serves as a platform for the real time monitoring of ankle sprain injury risk of the “sprain-free shoe”.

METHOD: Five male subjects (age = 21.8 $\pm$ 1.48 year, height = 1.7 $\pm$ 0.03 m, body mass = 66.2 $\pm$ 7.19 kg) with healthy ankles were recruited from the athletic team of The Chinese University of Hong Kong. The study was approved by the university ethics committee. For each subject, 3 trials on 5 different simulated supination sprain motions were performed on the supination sprain simulator (Chan et al., 2008). By rotating the fall platform of the supination sprain simulator, different degrees of supination from inversion to plantarflexion can be simulated. (when the fall platform set at 0^o was puer inversion, at 23^o, 45^o and 67^o were supination and 90^o was pure plantarflexion). The different type of supination sprain motions allowed a wide range of data to be collected. A uni-axial gyroscope (Sengital Ltd., Hong Kong, China) which measured heel tilting velocity was attached on the shoe surface at the heel position of the right shoe to collect the heel tilting velocity at a sampling rate of 500Hz. The axis of the gyroscope was aligned to measure the foot segment inversion and

eversion velocity. For the consistency of this alignment axis, all subjects wore the same shoe with the sensor attached throughout the study.
The heel tilting velocity was also obtained by an optical motion analysis system as a standard to validate the data obtained by gyroscope. Twelve reflective markers (5 mm in diameter) were attached to lateral fibula epicondyle, tibial tuberosity, lateral proximal shank, medial proximal shank, anterior distal shank, lateral distal shank, medial distal shank, posterior heel, lateral heel, medial heel, medial foot and dorsal foot, either on the skin or shoe surface. Marker coordinates were recorded by an optical motion analysis system with 16 cameras (VICON, UK) at 500Hz. The marker coordinates were filtered by Generalized Cross-Validation package of Woltring with 15Hz cut-off frequency (Woltring, 1986). A static calibration trial in the anatomical position served as the offset position to determine the segment embedded axes of the shank and foot segment. The foot and shank segment were embedded with the Laboratory Coordinate System (LCS). A singular value decomposition method was employed to calculate the transformation from triad reference frame to anatomical shank and foot reference frame (Soderkvist, 1993). Joint kinematics was deduced by the Joint Coordinate System (JCS) method (Grood, 1983). Foot tilting angle was defined as the angle between the LCS vertical axis and the normal of the foot transverse plane, and the foot tilting velocity is its change with respect to time. The data analysis was processed by a customized Matlab program.

RESULTS: Table 1 shows good agreement between the tilting velocity measured by gyroscope (gyro data) and optical motion analysis system (standard data). The average of intra class correlation was higher than 0.9, except in the case of pure plantarflexion (90^o). Root mean square error (RMSE) difference between the two methods was between 8.21 – 37.11 deg/s. Table 2 shows the peak magnitudes obtained by gyroscope and optical motion analysis system. The peak magnitudes of the gyro data were about 88 – 135% of that of the standard data except that in the case of plantarflexion. For the case of plantarflexion, the maximum difference was up to 200%. Figure 1 shows the pattern and the absolute error of the tilting velocity obtained by two methods of one selected trial in each supination angle. In general, the gyroscope data followed the pattern of the standard data well.

Table 1. Accuracy of the peak tilting velocity as represented by the intra class correlations (ICC) and the root mean square error (RMSE) in the unit of deg/s

Supination	0^o		23^o		45^o		67^o		90^o	
Angle	ICC	RMSE	ICC	RMSE	ICC	RMSE	ICC	RMSE	ICC	RMSE
Subject 1	0.81	24.59	0.99	9.29	0.96	11.76	0.94	14.60	0.89	13.03
Subject 2	0.97	37.11	0.93	9.63	0.97	10.66	0.83	11.27	0.74	11.32
Subject 3	0.88	18.75	0.95	16.37	0.99	10.59	0.94	12.17	0.79	13.37
Subject 4	0.97	18.30	0.94	13.40	0.98	10.47	0.97	8.21	0.70	21.13
Subject 5	0.95	13.51	0.99	24.55	0.96	11.04	0.87	20.36	0.81	28.46
Average	0.92	22.45	0.96	14.65	0.97	10.90	0.91	13.32	0.79	17.46
S.D.	0.07	9.09	0.03	6.25	0.01	0.52	0.06	4.55	0.07	7.22

Table 2. Peak magnitudes of tilting velocity obtained by gyroscope (gyro data) and optical motion analysis system (standard data) in the unit of deg/s

Supination Angle	0^o		23^o		45^o		67^o		90^o	
	Standard data	Gyro data	Standard data	Gyro data	Standard data	Gyro data	Standard data	Gyro data	Standard data	Gyro data
Subject 1	448.3	405.3	283.5	271.2	203.7	225.5	113.6	127.4	102.9	106.0
Subject 2	387.1	383.5	238.5	238.2	191.6	221.7	158.8	146.2	73.6	68.9
Subject 3	244.8	222.2	299.4	283.7	204.3	211.1	120.3	161.9	172.3	240.5
Subject 4	431.7	379.8	208.3	204.9	195.3	190.7	121.0	130.1	123.3	240.2
Subject 5	290.0	291.1	396.2	363.0	194.2	196.1	142.4	146.9	90.4	269.2

Average	306.4	336.4	285.2	272.2	197.8	209.0	131.2	142.5	112.5	185.0
S.D.	89.2	77.4	71.8	59.3	5.8	15.3	18.8	14.1	38.0	90.7

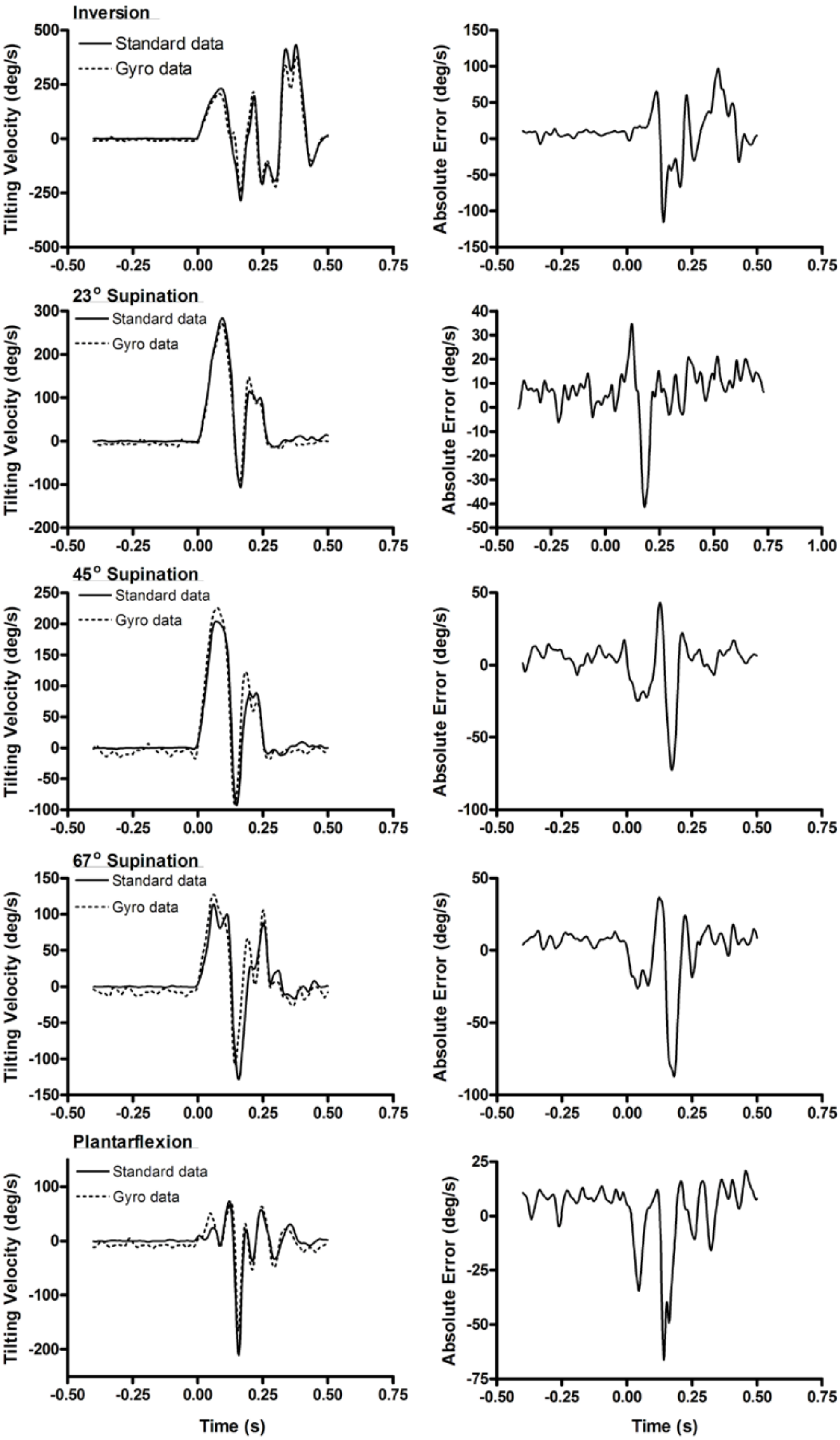

Figure 1. The pattern and the absolute error of the filtered tilting velocity obtained by gyroscope and optical motion analysis system of one selected trial in each supination angle.

DISCUSSION: In this study, the measurement of tilting velocity by a gyroscope was validated by an optical motion analysis system. The result shown that it have acceptable accuracy (ICC>0.9), except for the peak magnitudes of the tilting velocity in the case of plantarflexion (90^{o}), which with a relatively large variation. It should be noted that the value of the peak tilting velocity was not consistence between optical motion analysis and gyroscope. In some cases, the peak tilting velocity were higher with the gyroscope, in others they were lower. Such deviation may due to the variations in the distances between markers and also could be a result of camera noise, limited sight of markers, or vibrations of the marker (Ehara, 1997). Sensitivity and noise of the gyroscope is also sources of error. The limitations of this experiment include placing markers on shoe surface but not the foot and the use of sprain simulator but not the real motion. Simulated sprain motion was chosen in this study since we are developing a “sprain-free sport shoe” (Chan, 2006). We are seeking an inexpensive and simple way for real-time ankle sprain motion monitoring and detection. The present method only needs one gyroscope, thus is inexpensive and readily available to be implanted in a sport shoe. In the future, we will compare the foot tilting velocity data to the data collected from other common sporting motion, in order to establish a database to define the criteria for the identification of a hazardous ankle spraining motion.

CONCLUSION: This study showed the feasibility of using a gyroscope to monitor the foot segment tilting velocity. The device serves as a platform for a recently developed “sprain-free sport shoe” for real-time monitoring and detection of hazardous ankle spraining motion. Its advantage is being inexpensive and tiny, and could be implanted into sport shoe easily.

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THE EFFECT OF BODY MARKERS ON GOLF DRIVING PERFORMANCE

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KEYWORDS: accuracy, body markers, golf, launch characteristics, performance.

INTRODUCTION: No study to date has reported if and how the use of body markers used in three dimensional optical tracking methods to study swing kinematics in golf affect movement performance. Egret *et al.* (2004) studied the use of wired electromyographic equipment during the golf swing and concluded that the equipment significantly influenced the kinematic pattern of the golf swing. Researchers have previously concentrated their methodological analyses on such factors as the type of marker used, either wand or skin marker (Kirtley, 2002) or skin movement artefact during movement (Holden *et al*., 2007). The golf swing is a movement that is closed-chain, non-impact and does not cause excessive unwanted movement of skin and wand markers. It is therefore concluded that the golf swing lends itself well to kinematic analysis using body markers. The aim of the present study was to evaluate the effect of body markers on golf driving performance for tests carried out in a laboratory setting.

METHOD: Seven category 1 (<5 handicap) golfers (22.1 ± 2.3 yrs, 77.4 ± 9.7 kg, 1.80 ± 0.09 m and 0.2 ± 2.4 handicap) took part. All golfers were male and right-handed. Performance for each shot was determined through analysis of club head and ball impact characteristics measured using a commercially available launch monitor (GolfTek™ Pro V). Subjects were positioned on an artificial grass surface wearing golf spikes as they normally would on a golf course and selected their own tee height. Thirty four body markers were attached to the subject: acromion, lateral epicondyle of the elbow, wrist centre, C4, anterior superior iliac spine, sacrum, greater trochanter, lateral epicondyle of the knee, anterior epicondyle of the knee, medial malleolus, lateral malleolus, 2nd metatarsal head, heel, and the geometric centre of mass (COM) of the upper and lower arms, and upper and lower legs. Humeral and radial markers were positioned on 63.5 mm (2½″) wands and femoral and tibial markers were positioned on 101.6 mm (4″) wands. Additional club markers were placed on the golf club shaft 254 mm (10″) from the club butt and on the toe of the club head.

A 240 Hz 5-camera Motion Analysis Corporation™ Falcon Analogue system tracked all body and club markers during the subjects' swings when body markers were attached, and only the club markers for shots performed without body markers. Subjects warmed up as they normally would before playing golf. Using their own driver subjects were instructed to hit eight shots for each randomly assigned set-up along a target line marked on the floor into netting 4.5m away.

RESULTS AND DISCUSSION: Significant differences ($z = -2.521$, $p < 0.05$) were noted for ball velocity when shots were hit with and without markers (Table 1).

Table 1. Launch monitor data for the golf swing with and without body markers

Measure	With Body Markers	Without Body Markers
Club Head Velocity (ms^{-1})	49.96 ± 0.67	49.40 ± 1.07
Ball Velocity (ms^{-1})*	69.62 ± 0.85	66.70 ± 0.93
Club Head Orientation (˚)	1.25 ± 3.24	3.00 ± 0.93
Tempo (s)	0.82 ± 0.01	0.81 ± 0.02
Backspin (rev/min)*	2676.5 ± 312.2	3263.6 ± 672.1
Sidespin (rev/min)*	-493.1 ± 423.1	189.0 ± 701.5
Ball launch angle (˚)	11.13 ± 2.12	10.63 ± 1.88

**p< 0.05*

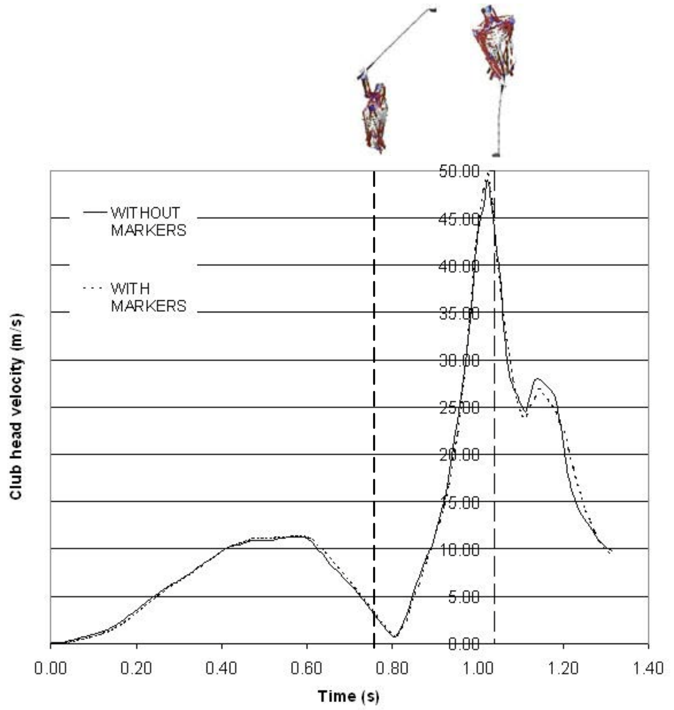

Figure 1. Club head velocity with and without markers

Shots taken without body markers averaged 2.92 ms^{-1} reduced ball velocity (-4.19%). Club head velocity did not prove significant with only 0.56 ms^{-1} difference (Figure 1). Both ball backspin and sidespin component showed significant differences ($z = -2.38$, $p < 0.05$). Swing tempo did not show differences between the two conditions. The present study illustrates that attachment of body markers that would normally be used to study the kinematics of the golf swing via passive marker based optical three dimensional systems, induces minor changes in the swing as inferred by a change in ball launch velocity and spin rates. Ball velocity normally indirectly correlates with carry distance, and sidespin component of flight normally directly correlates with shot accuracy. Sidespin, or non-horizontal component of the ball was shown to orientate left, or anti-clockwise for shots performed with markers attached. This would indicate that those shots performed with markers attached may have been less accurate, producing a more excessive right-to-left 'hook' shape. Subjects seemed to overcompensate when markers were attached, potentially sacrificing accuracy for power. Thus, important components of club head – ball impact which affect ball launch characteristics were altered.

CONCLUSION: Body markers significantly affect key shot performance measures, and field testing is required to ascertain accuracy and carry in a future study to support these findings. A number of the measures recorded, including club head velocity at impact, club head orientation, swing tempo, and ball launch angle, were relatively unaffected by the presence of body markers. Ecological validity is a concern during lab based experimentation and further investigation of the effect of experimental testing equipment on outcome performance, and shot accuracy on the golf course, is warranted for greater subject numbers.

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VALIDATION OF AN ELECTRONIC JUMP MAT

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The purpose of this investigation was to determine the validity of a commonly used Electronic Switch Mat (ESM), or jump mat, compared to force platform data. 10 subjects completed 3 Squat Jumps (SJ), 3 Countermovement (CMJ) jumps and 3 Drop Jumps (DJ) from a height of 30cm. The jumps were performed on a force platform (FP) with an ESM positioned on top of the platform. CMJ (r=0.996) and SJ (r=0.958) jump heights correlated very strongly with force platform data however drop jump data was not as strong (r=0.683). The ESM can accurately calculate CMJ height, and SJ height. However, the faster contact times, and rapid movements involved in a DJ may limit its reliability when giving measures of contact time, flight time, and height jumped for DJs.

KEYWORDS: force platform, jump mat, validation.

INTRODUCTION: Several methods of measurement exist for the vertical jump and its performance components, and numerous pieces of equipment are available to the biomechanist and strength and conditioning coach to obtain a measure of lower body power from their athletes. Klavora (2000) highlighted several field tests that are commonly used in research and training contexts and describes the methodology and relative advantages and disadvantages of using each test. The tests that are described include the jump and reach test, belt tests, and electronic switch mat (ESM) tests. Many coaches use ESM's to measure height jumped by their athletes due to the cost effectiveness and portability of such a device (Klavora, 2000; Flanagan & Comyns, 2008). However, the validity of the ESM has not been ascertained and widely published in relation to fixed force platform data. The purpose of this study was to determine the validity and reliability of a commonly used electronic jump mat against a ground mounted force platform for the purpose of measuring various parameters during three types of jump.

METHOD: 10 subjects were recruited (23.6 ± 2.2 yrs, height 174.11 ± 16.63 cm, mass 77.37 ± 16.63 kg) who were all familiar with the three types of jump performed: Squat jump (SJ), Countermovement jump (CMJ) and Drop jump (DJ). The subject base included three track & field athletes, two gaelic hurling players, two Olympic weightlifters, two recreational runners and one rugby union player. Subjects completed three squat jumps, three countermovement jumps and three drop jumps from a height of 30 cm. The jumps were performed on an AMTI OR6-5 force platform operating at 1000 Hz. The jump mat was positioned on top of the force platform and the platform reset. Measured parameters during this study included Flight Time (FT) and Height Jumped (HJ) for both the CMJ and SJ. For the DJ, Contact Time (CT) and Reactive Strength Index (RSI) were also measured in addition to FT and HJ.

For the purpose of the current study an electronic switch mat was used (FLS JumpMat, Tyrone, Ireland). The ESM instrument includes a square mat attached to a hand-held monitor (Figure 1). With the aid of micro switches embedded in the mat, flight time was measured as the interval between feet lift-off from the mat to landing again on the mat.

Subjects warmed-up as they normally would before vigorous activity which included 5 minutes gentle jogging followed by 20 minutes of dynamic progressive ballistic exercises, hops, bounds and jumps. Subjects were permitted 180 seconds between repeats of the same type of jump, and 300 seconds between sets of different jump types to avoid fatigue. Subjects wore shorts, t-shirt and comfortable running shoes. The order of jump type was randomly assigned. Data were tabulated from the AMTI force platform software 'Bioware' and recorded from the ESM output box (Figure 1) and were used to calculate common jump parameters as shown in Table 1.

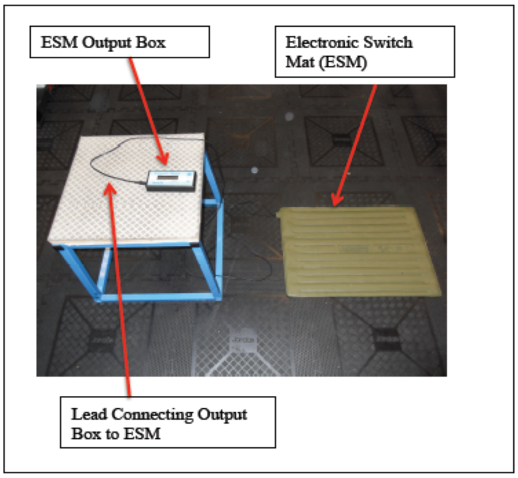

Figure 1. Electronic switch mat positioned over a force platform.

Table 1. ESM and force platform measures and calculations.

Measure	Description
Flight Time (FT)	Time between take-off and landing.
Contact Time (CT)	Time between the initial landing following a drop and the subsequent take-off point.
Height Jumped (HJ)	Calculated from the FT for each jump using the second mathematical equation of linear motion. $s = ut + ½at^2$, s = displacement u = initial velocity = 0 ms^{-1} t = time to top of jump a = acceleration due to gravity
Reactive Strength Index (RSI)	**RSI = HJ / CT** HJ = Height Jumped CT = Contact Time
Take-off Velocity (V)	**V = √2GHJ** G= acceleration due to gravity HJ = Height Jumped
Power (P)	**P = Vm** V= Take off velocity m = Mass of Subject
Time to Peak Ground Reaction Force (TPGRF)	Time taken for the subject to reach their maximum.

Correlation coefficients were used to identify the strength and directionality of the relationships between the various parameters taken by the force platform and ESM apparatus.

RESULTS: Table 1 highlights descriptive statistics for all parameters as measured by the ESM and the ground mounted force platform.

Table 1. Descriptive statistics for all parameters measured by the ESM and force platform.

	CMJ Height (cm)	CMJ Flight Time (sec)	SJ Height (cm)	SJ Flight Time (sec)	DJ Height (cm)	DJ Contact Time (sec)	DJ Flight Time (sec)	DJ RSI (no units)
ESM Max	0.50	0.64	0.51	0.65	0.34	0.92	0.53	2.45
ESM Min	0.20	0.40	0.17	0.37	0.14	0.14	0.33	0.64
ESM Average	0.32	0.51	0.31	0.49	0.22	0.24	0.42	1.12
ESM St Dev	0.09	0.07	0.09	0.08	0.05	0.15	0.05	0.49
FP Max	0.49	0.64	0.50	0.64	0.34	0.82	0.53	2.52
FP Min	0.19	0.39	0.16	0.36	0.13	0.12	0.32	0.42
FP Average	0.31	0.50	0.29	0.48	0.21	0.25	0.41	0.98
FP St Dev	0.09	0.07	0.09	0.08	0.06	0.13	0.06	0.47

In addition, Table 2 shows the Pearson's correlations between all parameters measured by the EMS and the force platform. All correlations were significant ($p>0.05$) and very strong, except for the DJ Contact Times, and DJ Flight Times.

Table 2. Pearson's correlation coefficients and RMSD for all parameters measured by the ESM and force platform.

	CMJ Height (FP)	CMJ Flight Time (FP)	SJ Height (FP)	SJ Flight Time (FP)	DJ Height (FP)	DJ Contact Time (FP)	DJ Flight Time (FP)	DJ RSI (FP)
Pearson's r	0.996	0.996	0.958	0.946	0.683	-0.173	0.269	0.938
RMSD	2.040	0.000	8.240	0.001	0.000	0.02	0.01	0.040
	CMJ Height (ESM)	CMJ Flight Time (ESM)	SJ Height (ESM)	SJ Flight Time (ESM)	DJ Height (ESM)	DJ Contact Time (ESM)	DJ Flight Time (ESM)	DJ RSI 2 (ESM)

Figure 2 illustrates the strength of the relationships for the key measure height jumped and Table 3 shows for this measure the magnitude by which ESM data is greater.

Table 3. Force platform jump height values as a percentage of ESM data.

	CMJ Height (%)	SJ Height (%)	DJ Height (%)
ESM	100	100	100
FP	96.23	98.04	97.96
% difference	3.77	1.96	2.04

DISCUSSION: Results comparing the two instruments showed very strong correlations for SJ height (0.985), SJ flight time (0.946), CMJ height (0.996), and CMJ flight time (0.996). Although the correlation between the two DJ heights were strong and significant (0.683), the correlations between contact time and flight time in the DJ was very weak (-0.173, and 0.269 respectively). Therefore, the ESM tested in this study is a valid instrument for measurements of height jumped and flight time in CMJ and SJ, but will not produce accurate results when measuring DJs. A possible explanation for the low correlations found for contact time and flight time in the DJs is that the DJ involves a much smaller ground contact period that the relatively inexpensive ESM cannot accurately detect. All data were in line with that reported by Bobbert *et al.* (1996). The use of RSI as a measure of feedback to the athlete and coach for DJ training is relatively new but has been shown here to be ascertained reliably via a jump mat (r=0.938).

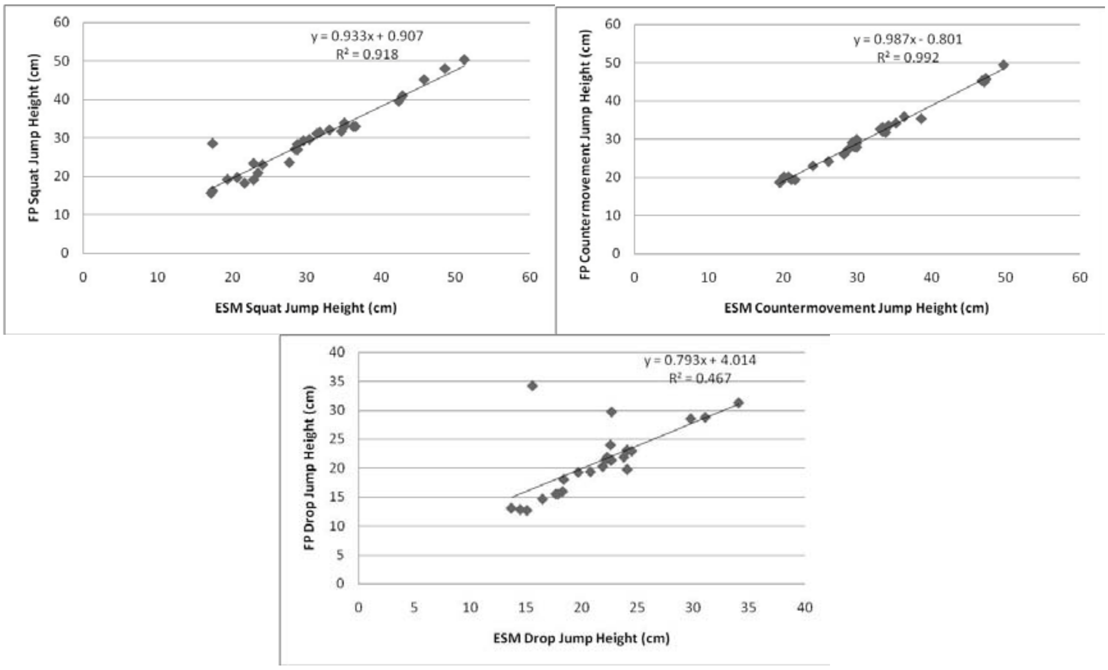

Figure 2. Regression analysis for ESM and force platform height jumped.

CONCLUSION:

- The jump mat consistently yielded data higher in magnitude than the force platform. 1.96%- 3.77% of a difference between FP and ESM readings for CMJ, SJ, and DJ results.
- Correlations between CMJ and SJ on the FP and ESM were all significantly correlated to a high degree (0.996, $p>0.05$).
- DJ data were also significantly correlated (0.683, $p<0.05$) although the correlations were weaker than observed in the CMJ and SJ.

The commonly used jump mat (FLS JumpMat) tested in the current study gave reliable results of vertical jump tests when compared against a ground mounted strain gauge force platform (AMTI OR6-5). This has implications for the strength and conditioning coach in that they can confidently test their athletes in the field, thus avoiding bringing the athletes to the laboratory when relatively simple measures such as height jumped, RSI, and contact time are required. The jump mat may also be a useful resource for researchers carrying out large scale investigations with high subject numbers because the ESM will save time in information processing and experimental set-up. However the ESM is not without its limitations, such as the reduced amount of data produced and errors associated with DJ protocols.

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CAN MUSCLE ACTIVATION BE INCREASED WHEN MODIFYING THE DUMBBELL CHEST PRESS? AN ELECTROMYOGRAPHIC COMPARISON

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KEY WORDS: unilateral, stability, bench press, core.

INTRODUCTION: The development of core stability is an important part of the athlete strength and conditioning process. Research has shown that when traditional resistance exercise is performed on a stability ball, core muscle surface electromyography (sEMG) is increased (Anderson & Behm, 2005). However, investigators have also demonstrated this can lead to a considerable reduction in agonist muscle sEMG (Behm & Anderson, 2006). It has been suggested that by replacing traditional bilateral resistance exercises with unilateral variations the reduction in agonist muscle sEMG associated with stability ball resistance exercise can be avoided (Behm & Anderson, 2006). However, this theory has not been tested on the chest press. Therefore, the aim of this study was to compare core and agonist sEMG during chest press performance to establish whether unilateral chest press exercise would increase core sEMG without compromising agonist sEMG.

METHODS: After gaining university ethical approval, six recreationally resistance trained males (mean ± SD age: 23 ± 1 years; mass: 79.8 ± 9.7 kg; height: 1.8 ± 0.01 m) were recruited and provided informed consent.

Two testing sessions were conducted. During the first, bilateral bench dumbbell chest press (Figure 1) maximal strength (one repetition maximum-1RM) was established using the methods proposed and used by Bompa and Cornacchia (1998). Seven days later subjects performed two sets of five repetitions with 60% 1RM in the four dumbbell chest press variations that are shown in Figure 1.

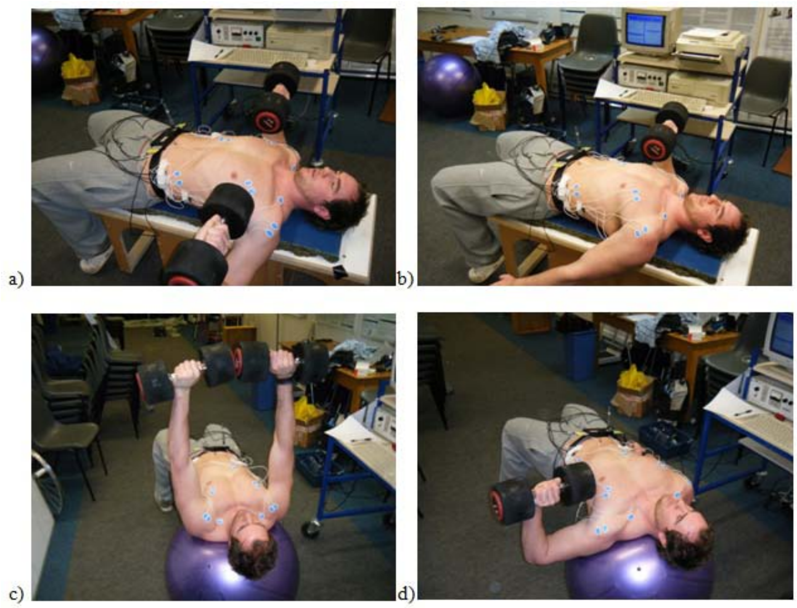

Figure 1. Demonstrations of the chest press variations that were studied: a) bilateral and b) unilateral bench chest press (CP) (L and R), and c) bilateral and d) unilateral ball CP (L and R).

All subjects rested for 3 minutes between each set and 10 minutes between each exercise. The sEMG of seven muscles (agonist: left and right anterior deltoid, pectoralis major; core: left and right oblique, and rectus abdominis were recorded at 500 Hz using a radio telemetry system (MIE Medical Research Ltd., Leeds, UK). Following skin preparation, AgAgCl surface electrodes were

placed in accordance with (Cram et al., 1998; Figure 1), with a maximal voluntary contraction (MVC) recorded against manual resistance for each muscle under investigation. Mean concentric phase muscle activity for each muscle during each repetition was calculated by summing integrated sEMG data. Repetition data was then averaged for further analysis.
All data were presented as mean (± SD). Differences between the different exercise core and agonist sEMG were established using repeated measures ANOVA and planned comparisons on SPSS version 16 for Windows (SPSS Inc., Chicago, USA). An alpha value of $p \leq 0.05$ set to establish statistical significance.

RESULTS AND DISCUSSION: Descriptive statistics are presented in Table 1. Exercise type significantly affected the right oblique sEMG (p<0.05, Table 1), but not the left oblique (p = 0.414), left (p = 0.402) and right (p = 0.255) deltoid, left (p = 0.188) and right (p = 0.214) pectoralis major, or rectus abdominis (p = 0.343, Table 1).

Table 1. Mean (± SD) sEMG (% MVC) during the different variations of the dumbbell chest press.

	Bench Bilateral	Bench Right	Bench Left	Ball Bilateral	Ball Right	Ball Left
Right deltoid	58 (12)	60 (12)	66 (5)	65 (10)	64 (12)	57 (3)
Left deltoid	52 (14)	58 (6)	61 (13)	56 (13)	60 (7)	61 (7)
Right pectoralis	58 (7)	57 (9)	38 (4)	47 (3)	57 (9)	51 (31)
Left pectoral	58 (6)	38 (8)	67 (9)	52 (8)	35 (.4)	61 (1)
Right oblique	59 (8)[a]	58 (3)	60 (20)	63 (8)[b]	53 (6)	77 (3)
Left oblique	58 (11)	60 (8)	61 (10)	52 (23)	60 (17)	78 (10)
Rec ab	36 (9)	40 (15)	40 (4)	33 (6)	45 (14)	41 (3)

Rec ab = rectus abdominis; a = 10.2% greater than unilateral ball exercise (p < 0.05); b = 15.9 % greater than unilateral ball exercise (p < 0.05).

Unilateral bench exercise did not elicit significantly greater sEMG than the unilateral ball equivalent. However, unilateral bench exercise demands on the rectus abdominis were between 10% (for bench exercise) and 23% (for ball exercise) greater than the bilateral equivalent, indicating that a considerable component of the core musculature had a greater demand made of it when performed unilaterally, especially when a stability ball was used. Further, unilateral bench exercise offered no clear benefit over the ball equivalent for the obliques. Although right oblique activity was 8.6% greater, left oblique activity was 28% less (Table 1). This is further supported when the agonist sEMG is reviewed. Deltoid activity tended to be greater during stability ball exercise. However these differences, in addition to the differences between the stability ball and bench exercise pectoralis activity were minimal. This suggests that unilateral stability ball chest press exercise places a greater demand on a considerable portion of the core musculature with little affect on the agonist musculature.

CONCLUSION: The results of this study refute the contention that unilateral variations of traditional chest press exercise elicit greater demands on core and agonist sEMG. Greater demands are placed on these muscle groups during unilateral stability ball chest press exercise.

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APPLICABILITY OF A FULL BODY INERTIAL MEASUREMENT SYSTEM FOR KINEMATIC ANALYSIS OF THE DISCUS THROW

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The aim of the study was the application of a full body inertial measurement system (IMS) for a kinematic analysis of the discus throw and the evaluation of its applicability. For this purpose, one male sports student performed three discus throws equipped with the IMS. All trials were additionally filmed by high-speed video. The results indicate that performance-relevant information can be obtained regarding the temporal coordination of the body segments and body joint angles. Limitations exist for the accurate detection of the last foot contact related instant and the discus release instant by solely using the IMS data.

KEY WORDS: motion analysis, inertial sensors, discus throw.

INTRODUCTION: The discus throw can be divided into two stages, the launch and the flight (Hubbard, 1989). The launch phase consists of the thrower's movements in the circle, aiming at optimizing the release of the discus in order to maximize the throwing distance. The most relevant discus release parameters are the release speed, the release height and the release angle (Bartlett, 1992). To achieve maximum release speed through complex rotational movements, optimizing technique is vital in discus throwing (Leigh & Yu, 2007). Therefore, biomechanical movement analyses are performed to evaluate techniques of elite athletes on the basis of video information (Dickwach & Knoll, 2003; Hildebrand & Perlt, 2007; Leigh & Yu, 2007). Problems arising from the extraction of three-dimensional (3D) kinematic information on the basis of two-dimensional (2D) video data can be associated with accuracy of position data (in particular for rotational movements), temporal resolution (in general 50 or 60 Hz) and the time needed for subsequent analysis. Therefore, alternative methods overcoming these problems should be taken into consideration.

Advances and miniaturization in sensor technology enabled ambulatory motion analysis with the use of inertial sensors (e.g. Bonato, 2003). By combining inertial sensors (accelerometers and gyroscopes) and magnetometers, fixed to body segments, drift free orientation of the segment can be estimated. With the integrated use of a biomechanical model and sensor fusion algorithms, full 3D kinematics of the body segments can be obtained (Roetenberg et al., 2009). A full body inertial measurement system (IMS), therefore, implies potential for the analysis of complex movements in a real sports environment. Recent applications included motion analysis in alpine skiing (Brodie et al., 2008) and freestyle snowboarding (Krüger & Edelmann-Nusser, 2009). Krüger & Edelmann-Nusser (2010) also reported good agreement between IMS and a video based system for analyzed knee angles in alpine skiing.

The purpose of this pilot study was the application of an IMS for a kinematic analysis of the discus throw (characterized as a complex rotational movement) and the evaluation of its applicability in terms of: 1. What kinematic data relevant to performance can be obtained? 2. How reliable is the detection of the critical instants of the discus throw (to subdivide the movement into phases) by solely using the IMS data?

METHOD: One male sports student (22 yrs, 1.88 m, 84 kg, former decathlete, 15 years athletics training experience, personal best discus performance: 48 m) performed three discus throws (indoors; 1 kg discus). The throws were simultaneously recorded using an IMS and high-speed video (HS). The IMS (MVN; Xsens Technologies, Netherlands) consists of a suit equipped with 17 inertial sensor units (MTx) and two transmission units. Each unit integrates 3D linear accelerometers, 3D rate gyroscopes and 3D magnetometers. The full scale range of all inertial sensors is 1200 deg/s (gyroscopes) and 50 m/s^2 (accelerometers), except for two sensors located at the lower arm and hand segment of the throwing arm with a measurement range of 100 m/s^2. IMS data acquisition was performed with the MVN Studio software (v2.6). As the input for the biomechanical model (23 segments, 22 joints) subjects' anthropometric data (9 parameters) were used. Prior to the discus throws the subject needed to perform four standard poses for the calibration of the IMS. All IMS data of the discus throws were then sampled at a rate of 120 Hz. The HS videos were recorded at a frame rate of 300 Hz (Exilim EX-F1, Casio; 512 x 384 pixels) from lateral perspective. Systems were time synchronized by detecting the instant of landing after a vertical jump in both the sensor and video data. 3D Segment orientations, angular velocities and accelerations as well as joint positions, velocities and accelerations were calculated in MVN Studio and exported for further processing using MATLAB (The Mathworks). Additional kinematic parameters that have been identified to be performance-relevant (Dickwach & Knoll, 2003; Leigh & Yu, 2007) were then computed and are presented in Table 1.

Table 1. Computed kinematic parameters

Parameter	Description	Unit
v_hip/ v_shoulder/ v_elbow/ v_hand	Resulting velocities/ accelerations for the hip/ shoulder/ elbow and wrist joint of the throwing arm	m/s
a_hip/ a_shoulder/ a_elbow/ a_hand		m/s^2
s_hand	Path of the wrist joint of the throwing arm	m
hip-shoulder	Separation angle of the hip and shoulder line	°
trunk tilt	Tilt angle of the trunk expressed as forward and backward tilt	°
knee left/ knee right	Angles of the knee joint with respect to the mediolateral axis	°

From the HS video data, the six critical instants (T1-T6) of the discus throwing movement were detected (see Figure 1), which lead to five phases: preparation, entry, airborne, transition, and delivery (Bartlett, 1992). The detection of the same instants by solely using the IMS data was done using the time course of the resulting velocity of the joint origin of the segment right hand for T1 (local minimum) and T6 (global maximum) and the ground contacts of the respective foot identified in the MVN Studio software (T2-T5).

Figure 1. Subject performing a discus throw wearing the IMS. Phases of the throw were subdivided according to six critical instants (from left to right): T1: maximum backswing; T2: right foot off; T3: left foot off; T4: right foot down; T5: left foot down; T6: release (Bartlett, 1992)

RESULTS: The velocity profiles of the throwing arm joints are presented in Figure 2. The results illustrate the temporal deviation of the velocity peaks within the last phase of the movement in order to accelerate the discus. Figure 3 shows the time course of relevant body angles, from which the particular values at critical instants can be extracted (e.g. hip-shoulder separation at release).

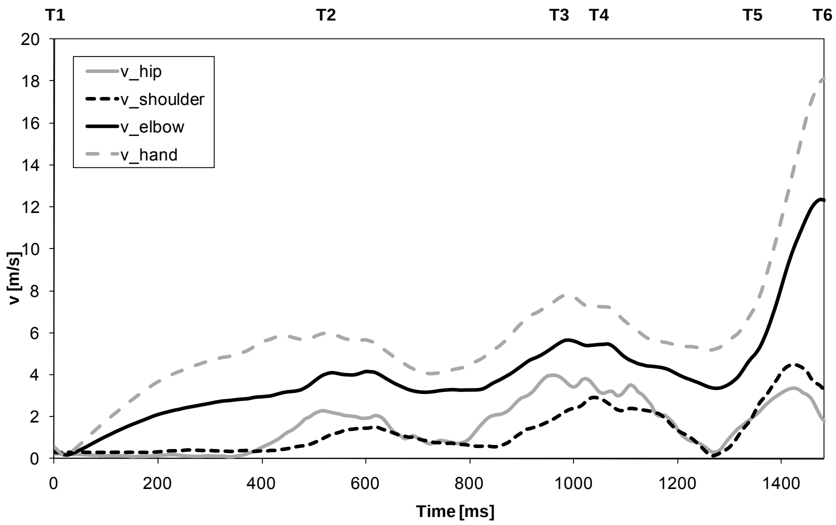

Figure 2. Resulting velocity of the respective joints of the throwing arm for one discus throw. T1 – T6 indicate instants of the movement phases.

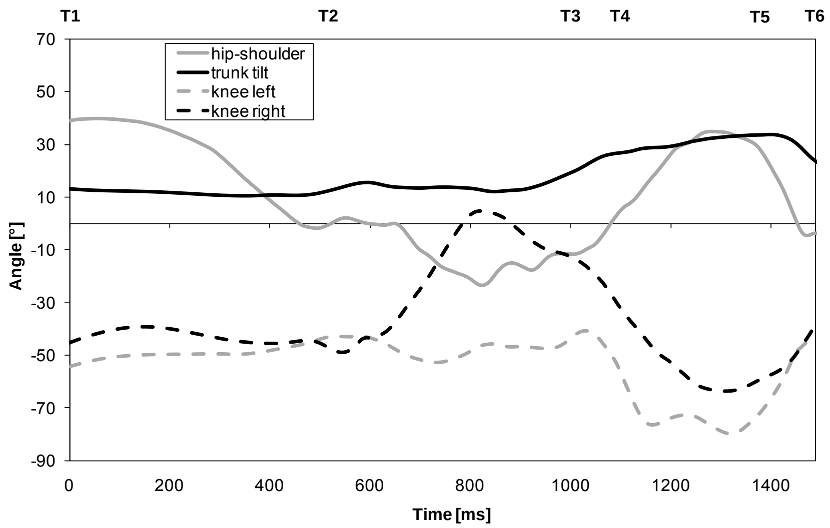

Figure 3. Angles of the left and right knee joint (<0°: flexion), hip-shoulder separation (>0°: line of the hips leading the line of the shoulders) and trunk tilt (>0°: forward tilt) for one discus throw. T1 – T6 indicate instants of the movement phases.

With respect to the subdivision of the movement phases of the discus throws Table 2 shows the phase durations observed and the differences when detecting the critical instants by solely using the IMS data.

Table 2. Phase durations determined from the HS videos (T1-T2: preparation; T2-T3: entry; T3-T4: airborne; T4-T5: transition; T5-T6: delivery) and time differences for each instant as detected by solely using the IMS data (n = 3 discus throws)

Instant	T1	T2	T3	T4	T5	T6
Phase duration (HS) [ms]	0	520-560	450-490	70-100	250-300	120-180
Diff. (IMS - HS) [ms]	42-75	0-8	0-2	8-50	167-308	17-25

DISCUSSION: Quantitative comparison of the obtained velocity profiles is limited, since no high-level athlete was investigated in this study and published values focus on discus release velocities (e.g. 23-25 m/s; Bartlett, 1992; Dickwach & Knoll, 2003). Moreover, they cannot be directly compared to the velocities at the wrist joint in this study (16-18 m/s). The profiles of specific performance-relevant body angles provide additional information on individual throwing technique. Leigh & Yu (2007) associated hip-shoulder separation angles at left foot off (25 to 50°), at left foot down (45 to 60°) and at the release (-10 to 0°) as well as the trunk tilt at right foot off (10 to 20°) and at the release (-10 to 0°) with throwing performance of female and male elite athletes. Dickwach & Knoll (2003) emphasized the relevance of the time courses of the knee angles that showed motion ranges of 25° (right knee) and 50° (left knee) for two elite athletes. Besides the reported parameters in the literature, a number of additional kinematic parameters can be easily extracted from the IMS data and should be analyzed for their particular performance-relevance. For the detection of the critical instants of the movement, sufficient accuracy can be achieved for the first three foot contact related instants, but not for the last one and also not for the instant of the discus release. Therefore, and since the relevant release parameters of the discus cannot be directly assessed by the IMS, additional video information would be necessary. Opposite to the advantages of the IMS for motion analysis in terms of an unrestricted capture volume and short-time data acquisition and analysis, possible drawbacks of the system, like an interference with the normal movement when wearing the IMS was not judged critical by the investigated subject.

CONCLUSION: This pilot study used a full body inertial measurement system for the kinematic analysis of the discus throw and emphasizes the potential of this approach for analysis of complex movements in sports. Like the discus throw, valuable application of the system is also expected for the rotational movements performed in shot put or hammer throw. Future studies need to evaluate the accuracy of the system for the assessment of kinematic data in particular in high-level athletes and should evaluate a possible interference with the movement.

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FACTORS DETERMINING THE SPIN AXIS OF A PITCHED FASTBALL

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KEYWORDS: baseball, pitching, spin.

INTRODUCTION: In order to make hitting more difficult for a batter, the pitcher introduces spin and alters the trajectory from a simple parabolic trajectory to one in which the aerodynamic force plays a significant role (Alaways and Hubbard, 2001). Although the direction of the spin axis greatly affects the lift force acting on the ball (Jinji and Sakurai, 2006), no study has been reported on how the spin axis of a pitched ball is determined in pitching motion. The purpose of the present study was to investigate the factors that determine the direction of the spin axis of the baseball. It is expected that movement of the upper extremity relates to the direction of the ball spin axis (Chin et al., 2009). The present study focused on the orientation of a pitching hand in a global coordinate system.

METHOD: Nineteen male baseball pitchers (height 176.2 ± 4.3 cm, mass 71.0 ± 7.4 kg) volunteered to participate. Fifteen subjects were right-handed, and four subjects were left-handed. All pitchers were regarded as over-hand or three-quarter-hand pitchers. Each pitcher performed pitching in an indoor pitching mound, pitched five to eight fastballs to a catcher 18.44 m away. A VICON MX motion analysis system (Oxford Metrics Inc.) was used to record the movements of the reflective markers attached to the pitcher's body (13 points) and the balls (4 points). The system operated 10 cameras at a sampling frequency of 1000 Hz.

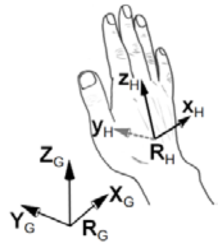

Figure 1. Definition of the global and local reference frames. R_G denotes the global reference frame; R_H, the hand reference frame.

For each pitcher, the trial in which the velocity of the pitched ball was found fastest was selected for analysis. Cardan rotation angles were used to define the orientation of the hand segment. The angles of the hand direction are designated right rotation/left rotation for the first rotation about the z_H axis, right sideways/left sideways for the second rotation about the y_H' axis, and backward tilt/forward tilt for the third rotation about the x_H'' axis (Fig. 1). For the analysis of the left-handed pitchers, a left-hand coordinate system was defined, with the Y_G and Z_G axes being the same as those of the right-hand coordinate system and the X_G axis directed opposite to that of the right-hand coordinate system.

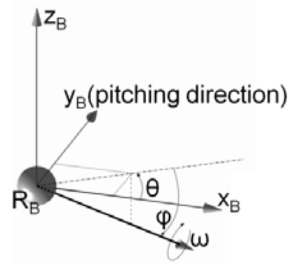

Figure 2. Definition of spin axis. The orientation of the angular velocity vector ω is specified by two angles φ and θ.

The angular velocity vector of the ball spin immediately after ball release (BRL) was calculated from the four hemispherical reflective markers attached to the pitched baseball; the calculation was performed with the method described by Jinji and Sakurai (2006). The direction of the angular velocity vector was expressed in the global reference frame, defined by the azimuth θ (the angle between X_B and the projection of the spin axis in the horizontal plane) and the elevation φ (the angle between the spin axis and the horizontal plane).

Pearson's product moment correlation coefficients were calculated to determine the relationship between the direction of the spin axis and the parameters representing the angles of the hand direction. The level of significance was set at 0.05.

RESULTS: The ball speed for the nineteen subjects was 34.0 ± 3.1 m/s (average ± S.D.). The spin rate was 27.4 ± 3.4 rps. The values of the angles θ and φ were 34.9 ± 14.1° and –28.4 ± 9.8°, respectively.

The hand was in the position of right rotation (18.2 ± 12.1°) at the maximum external rotation (MER) of the shoulder. The angle of right rotation attained the peak value at 12 ms before the BRL. Subsequently, the hand rotated to the left, and the ball was released with a right rotation (7.0 ± 12.1°). Further, the hand was in a position of left sideways rotation (–25.1 ± 10.2°) at the MER. Then it tilted to the right, and attainded the maximum value of right sideways rotation (41.6 ± 15.1°) at about time of BRL. In addition, the hand was in a position of backward tilt (107.8 ± 9.2°) at the MER. Then, it gradually tilted forward, and subsequently, the ball was released with a slightly backward tilt position (10.9 ± 11.4°) of the hand.
The angle of the spin axis θ exhibited the highest correlation with the angle of hand rotation at 6 ms before the BRL (r = 0.840, Fig.3). In addition, the angle of the spin axis φ exhibited the highest correlation with the angle of the hand's sideways rotation at 7 ms before the BRL (r = –0.725, Fig.4).

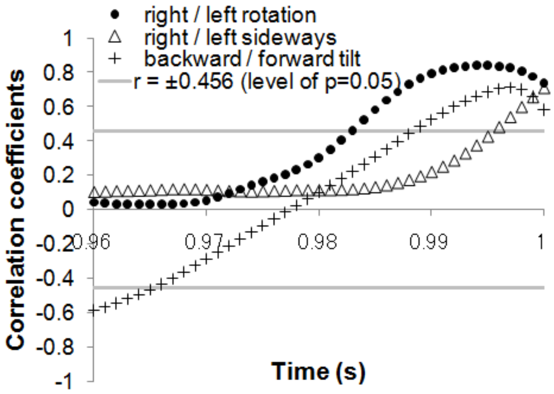

Fig.3: Changes in correlation coefficients between the direction of spin axis θ and orientations of the hand. The time of ball release is assigned t= 1.000 s.

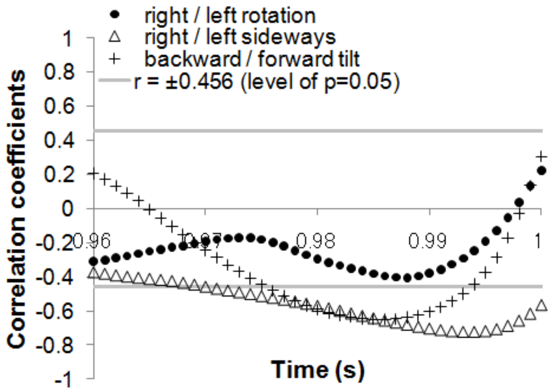

Fig.4: Changes in correlation coefficients between the direction of spin axis φ and orientations of the hand. The time of ball release is assigned t = 1.000 s.

DISCUSSION: Stevenson (1985) reported that the thumb of the throwing hand came off the ball at approximately 6 ms before the BRL. This instant of time is almost identical to the one when the angles of the spin axis have the highest correlation with the orientation of the hand in this study. As soon as the hand is partially open, the ball starts to roll or slide relative to the hand (Hore et al., 1996). Then, the ball rotates on the plane that is formed by the palm and fingers and is released from the fingertip. Therefore, the spin axis of the ball is parallel to the plane. It was found that the spin axis of a fastball was determined by these mechanisms.
The lift force of the pitched baseball is largest when the angular and translational velocity vectors of the ball are mutually perpendicular. In order to increase the lift force, the palm needs to face toward the home plate. On the other hand, there is a possibility of many types of balls with various spin axis directions and rotation speeds. Basically, it is expected that the direction of the spin axis is determined by the direction of the palm.

CONCLUSION: The orientation of the hand just before ball release was a significant factor in determining the direction of the spin axis.

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COMPARISON OF PELVIS KINEMATICS DURING THE BASEBALL PITCH: FATIGUED AND NON-FATIGUED CONDITIONS

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The purpose of this study was to identify changes that occur in pelvis kinematics as baseball pitchers fatigue during extended performances. Kinematic data describing the actions of the pelvis were collected using electromagnetic tracking techniques and calculated using the ISB recommendations. There were significant differences between non-fatigued and fatigued conditions in the angle of lateral pelvis flexion at maximum external rotation and release ($p<0.05$). As for the rate of axial pelvis rotation, no differences were observed between the non-fatigued and fatigued states. These results indicate that fatigue may play a major role in pitchers altering the actions of the pelvis during pitching.

KEYWORDS: Pitching, kinematics, fatigue

INTRODUCTION: Baseball pitching is often considered the most dynamic overhand movement in sports. Its violent nature repeatedly subjects the body to high magnitudes of joint kinetics (Adams, 1991). It is currently thought that the repeated nature of these stresses may be related to the high incidence of shoulder injury observed in baseball pitchers. From a biomechanical perspective, the manner in which each segment involved in pitching allows for optimal momentum transfer through larger segments to the smaller segments is commonly referred to as the kinetic chain (Kibler, 1991). In order for this momentum transfer to be optimized, the actions of the trunk and pelvis must be exceedingly efficient. Although it has become evident that control of the pelvis plays a major role in both performance and injury prevention (Werner et al., 2001; Aguinaldo et al., 2007; McKenzie, 2008), there are no known studies investigating how the kinematics of the pelvis are altered as pitchers fatigue. Therefore it was the purpose of this study to compare pelvic kinematics in baseball pitchers, while throwing the fastball, during non-fatigued and fatigued conditions.

METHODS: Ten male baseball pitchers (17.4yrs ± 3.27 yrs, 76.9 kg ± 12.2 kg, 178.2cm ± 7.2 cm) volunteered to participate. All subjects had recently finished their collegiate fall baseball season, and were deemed free of injury. Throwing arm dominance was not a factor contributing to participant selection or exclusion for this study.
Kinematic data were collected using The MotionMonitor™ electromagnetic tracking system (Innovative Sports Training, Chicago IL). Subjects had a series of electromagnetic sensors attached to the medial aspect of the torso and pelvis at the C7 and S1 locations respectively (Pope & Panjabi, 1985), as well as the distal/posterior aspect of the throwing humerus (Figure 1). To determine the instant of foot contact, a Bertec 40x60 cm force plate (Columbus, OH) was set to collect ground reaction forces at a rate of 1000 Hz. Sensors were affixed using double sided tape and then wrapped using flexible hypoallergenic athletic tape. Following the attachment of the electromagnetic sensors, a fourth sensor was attached to a stylus and used to digitize the palpated position of various bony landmarks (Myers et al., 2005). To accurately digitize selected bony landmarks, subjects stood in the neutral anatomical position while digitization was being completed.
After all sensors were attached subjects completed their own pre-competition warm-up. After subjects completed their warm-up, five maximal effort fastballs were thrown to a catcher located regulation distance from the pitching mound (18.44m). The pitching surface was positioned so that the subject's stride foot would land on top of the 40 x 60 cm Bertec force plate (Bertec Corp, Columbus, Ohio) which was anchored into the floor. After five fastballs for strikes were

thrown, subjects threw a 2kg ball into a portable rebounder until they reported their maximum perceived fatigue on a scale of 0 to 3. Once the subjects reported maximum perceived fatigue, they returned to the mound to throw five additional fastballs for strikes. Because preliminary data analysis indicated that current pitchers were remarkably consistent with their mechanics (low intra-subject variability), only those data from the fastest pitch deemed strike for non-fatigued and fatigued conditions were selected for analysis.

Throwing kinematics for right handed subjects were calculated using the standards and conventions for reporting joint motion recommended by the International Shoulder Group of the International Society of Biomechanics (Wu et al., 2002; Wu et al., 2005). Briefly, raw data regarding sensor orientation and position were transformed to locally based coordinate systems for each of the respective body segments. Euler angle decomposition sequences were used to describe both the position and orientation of both the pelvis and trunk relative to the global coordinate system (Wu et al., 2002; Wu et al., 2005). The use of these rotational sequences allowed the data to be described in a manner that most closely represented the clinical definitions for the movements reported (Myers et al., 2005).

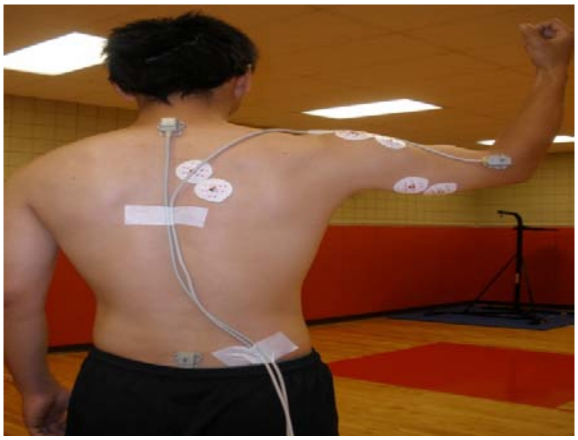

Figure 1. Sensor attachment for the humerus and trunk in the current study.

Throwing kinematics for left handed subjects were calculated using the same conventions; however, it was necessary to mirror the world z axis so that all movements could be calculated, analyzed, and described from a right hand point of view (Wu et al., 2002; Wu et al., 2005). Pitch velocity was also measured using a standard calibrated radar gun (Jugs, Tualatin, OR).

For each subject, means and standard deviations were calculated for each pelvis parameter. Prior to testing for mean differences the nature of the distribution was analyzed, and after the data were deemed to be normally distributed paired sample *t*-test were used to compare mean values between the non-fatigued and fatigued trials at following intervals: 1) stride foot contact (FC); 2) maximum shoulder external rotation (MER); 3) ball release (REL); and 4) maximum shoulder internal rotation (MIR). For each of the analyses, age was the independent variable and the kinematic parameter being analyzed was the dependent variable. Because the data were analyzed at four independent intervals, the level of significance for kinematic data was adjusted and set at alpha = 0.01. For additional information, the difference in pitch velocity was also tested using the same techniques.

RESULTS and DISCUSSION: The results of kinematic analyses are shown in Table 1. Of the 12 pelvis parameters analyzed in the current study, 2 were found to differ significantly between the non-fatigued and fatigued conditions. It has been previously suggested that throughout the pitching motion, kinematic alterations in the actions of proximal segments may result in kinematic alteration in the actions of distal segments (Putnam, C.A., 1993; Aguinaldo et al., 2007). Based on this logic, the actions of the pelvis could alter the actions of each subsequent segments involved in the pitching motion. The difference in the angle of lateral pelvic tilt at both MER and REL may result in an increase in the linear distance between the throwing hand and body. Previous investigations indicate that maximum elongation of the distance between the body and hand takes place during the arm cocking phase (Braun et al., 2009). This elongation occurs even when the vertical axis of the torso remains constant and may be exacerbated as the angle of lateral pelvis tilt increases in the fatigued state. As this distance increases at MER, there would be a resulting increase in the angle of external rotation. The combination of these actions could result in an increase in the magnitude of compressive force necessary to stabilize the shoulder joint.

Table 1. Kinematic differences between pitchers in the fatigued and non-fatigued states at specific instances throughout the pitching motion (Group mean ± SD)

	Non-fatigue (n=9)	Fatigue (n=9)	Sig.
FC			
Pelvis Posterior Tilt (^{o})	1.8 ± 4.1	1.7 ± 3.6	
Pelvis Leftward Tilt (^{o})	-4.4 ± 11.6	-4.2 ± 11.1	
Pelvis Leftward Rotation (^{o}/s)	307.5 ± 89.5	287.6 ± 57.9	
MER			
Pelvis Anterior Tilt (^{o})	-9.8 ± 14.9	-10.4 ± 11.3	
Pelvis Leftward Tilt (^{o})	-10.8 ± 11.8	-14.8 ± 11.3	*
Pelvis Leftward Rotation (^{o}/s)	268.8 ± 68.7	202.9 ± 54.3	
REL			
Pelvis Anterior Tilt (^{o})	-12.04 ± 14.29	-11.58 ± 12.81	
Pelvis Leftward Tilt (^{o})	-3.36 ± 5.24	-6.82 ± 3.87	*
Pelvis Leftward Rotation (^{o}/s)	167.8 ± 63.3	168.1 ± 53.4	
MIR			
Pelvis Anterior Tilt (^{o})	-8.37 ± 5.18	-7.62 ± 6.81	
Pelvis Leftward Tilt (^{o})	-1.18 ± 2.86	-2.69 ± 3.78	
Pelvis Leftward Rotation (^{o}/s)	136.2 ± 61.0	133.8 ± 50.4	

Note:

Following data collection sessions, it was determined that one subject supplied incorrect information on the medical history. Therefore, those data were removed from the analysis. * indicates a significant difference between fatigue levels ($p < 0.05$).

Additionally, as pitchers fatigue, there is a decrease in the rate of axial pelvis rotation. This decrease in the rate of pelvis rotation may result in pitchers transferring less angular momentum through the remaining body segments used in pitching. As the magnitude of this transferred momentum decreases, there may be a corresponding loss of pitch velocity. In the current study, it was observed that pitchers experienced a relative loss of pitch velocity (75 mph in non-fatigued state; 72 mph in fatigued state) once deemed fatigued. Although this difference was not significant, it should be considered that pitchers may alter the actions other segments in an attempt to make-up for the loss of pitch velocity.

CONCLUSION: It appears pelvis kinematics are altered as a baseball pitcher fatigues. Functionally, alterations in pelvic kinematics would affect the core or lumbopelvic-hip complex. It is the lumbopelvic-hip complex that essentially supports the torso and allows proximal stability for distal mobility (Kibler, 1991; Putnam, 1993). In addition, the lumbopelvic-hip complex is the fundamental link for efficient energy transfer from the lower extremity to the upper extremity. As pitchers fatigue, alterations in the actions of the pelvis may result in an increase in the stresses experienced by the body. When these stresses are increased, their cumulative effect on the structural integrity of the joints may be compromised. However, these results should be interpreted with caution as this study incorporated a relatively small sample from a pool of convenient subjects. Continued testing is needed to determine if these results can be observed in a larger, more representative sample.

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CHANGES IN LOWER LIMB JOINT RANGE OF MOTION ON COUNTERMOVEMENT VERTICAL JUMPING

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KEYWORDS: coordination, countermovement, range of motion, relationship.

INTRODUCTION: Humans naturally perform a countermovement (CM) action in order to attain a certain height or distance in jumping. This CM action initiates many movement coordination principles such as increasing lower limb joint range of motion (ROM) to allow the performer to be in a more effective position for jumping. In a study where the knee joint ROM was experimentally controlled Moran and Wallace (2007) reported a 17% increase in jump height with a 20° increase in knee joint flexion. Although Moran and Wallace (2007) experimentally controlled two knee flexion ranges, their results revealed a natural increase in hip joint ROM with subsequent increased knee flexion. No study, to the author's knowledge, has managed to identify the possible relationship between knee and hip joint ROM during a CM jump. The aim of this study is to examine the relationship between hip and knee joint ROM during CM vertical jumping and to determine if there are optimum joint flexion ranges at the hip and knee for jump height.

METHOD: Six healthy male participants (23 $\pm$ 6 years, 1.80 $\pm$ 0.10 m and 68.82 $\pm$ 8.1 kg) were asked to perform a series of 8 (non-arm) maximum effort CM vertical jumps with a wide array of peak knee flexion ranges. Verbal feedback and demonstrations on the required knee flexion range were given to each participant before commencement of each trial. Motion capture data collected at 240 Hz recorded a 6 degrees of freedom (DoF) lower limb retro reflective marker set from a standing position to the apex of the jump. Joint kinematics (sagittal plane) of the hip, knee and ankle were processed and analysed. Jump height was defined by the vertical displacement of the sacrum marker from standing to the apex of the jump. The coefficient of determination (R^2) was calculated to assess the relationship between kinematic variables during the CM action.

RESULTS: A strong positive curvilinear relationship was observed between peak hip and knee joint flexion ranges (R^2 = 0.827). Figure 1 illustrates that as peak knee flexion increased from approximately 60° to 110°, peak hip flexion was shown to increase linearly from 50° to 105° and as knee flexion further increased from approximately 110° and 150°, hip flexion subsquently plateaued between 110° to 120°. A moderate positive correlation (R^2 = 0.3462) was seen between peak hip flexion and jump height. Participants were shown to jump highest between peak hip flexion ranges of 100° to 130°.

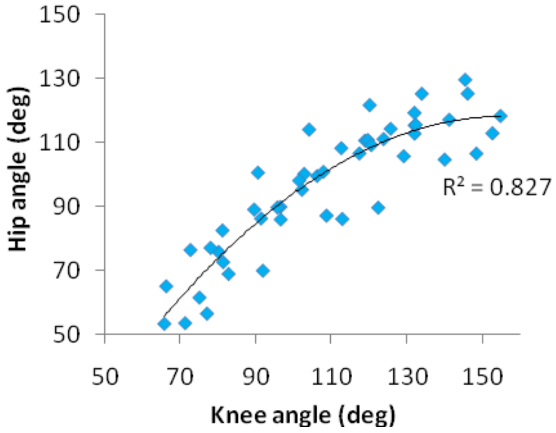
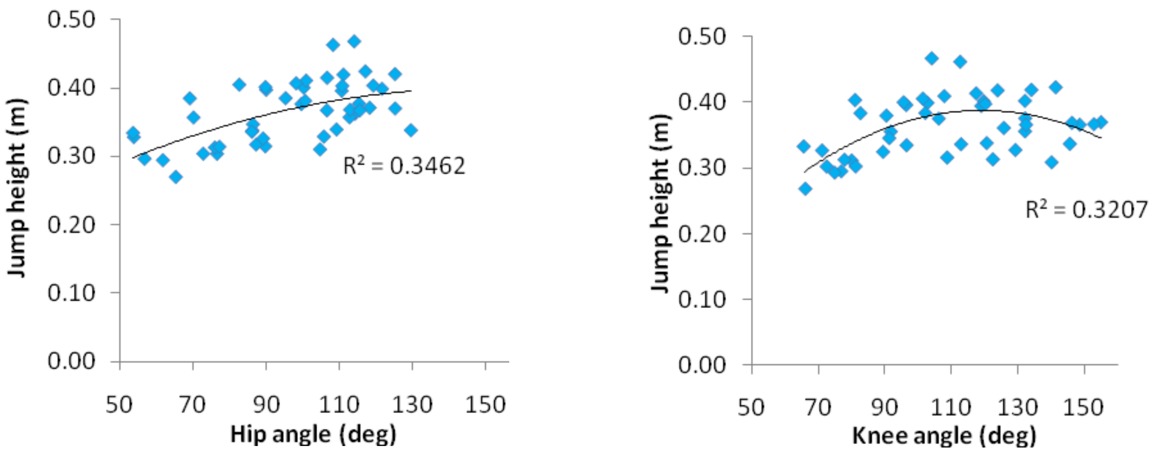


Figure 1. Peak flexion ranges of the hip and knee joints. Standing position was defined as 0°. A 2nd order polynomial trendline was used.



(a) **(b)**

Figure 2. Relationship between jump height and peak hip (a) and knee (b) flexion angles. Standing position was defined as 0°. A 2nd order polynomial trendline was used.

A moderate positive correlation (R^2= 0.3207) was also seen in peak knee flexion angle with jump height. The peak knee flexion ranges for optimal jump heights were shown to be between 100° to 130°, then after 130° jump height decreased.

DISCUSSION: A strong positive relationship between hip and knee joint ROM was clearly shown in CM vertical jumping. This movement coordination could be related to the muscle system (bi-articular muscles crossing both hip and knee joints) producing a constant force-length relationship (Voigt et al. 1995), thus as a result contributing to an effective power transfer for centre of mass (COM) vertical translation (Jacobs et al. 1996). Additionally, observations in this study suggest that during the deep CM (high peak knee flexion) the natural increase in peak hip flexion was used to keep the COM balanced and the resultant ground reaction force vector in front of COM in order to allow the lower limbs to optimally rotate forward during propulsion phase. A decrease in jump height when peak knee flexion was shown to be greater than 130° could be related to participants lack of practice and coordination on performing very deep CM as Domire and Challis, (2007) study suggested that in deep squat positions sub-optimal muscle coordination conditions arose from participants lack of practice.

CONCLUSION: The strong relationship between knee and hip joint ROM may be related to natural interlimb co-ordination of the human system to allow for optimal muscle contractile conditions. This finding suggests jumpers naturally engage both knee and hips joints in order to achieve an effective vertical jump height performance. The moderate relationships seen in hip and knee joint flexion ranges with vertical jump height may be accounted for individual jumping strategies and participants varying jumping ability.

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KINEMATIC ANALYSES IN TAEKWONDO POWER BREAKING MOVEMENT OF 360° JUMP BACK KICK

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The purpose of this study was to investigate the kinematic feature of the attack leg in 360° jump back kick. Eight Taekwondo coaches (Height: 175.7±5.7cm; Weight: 75.0±7.9kg; Age: 28.0±4.2years) performed the kick movement to break one inch board. Kinematic data was collected by ten Vicon cameras (250Hz). All variables were calculated by custom written program of MatLab. The results showed that the attack leg did not follow kinetic chain; the center of mass (COM) displacement in the lateral and vertical direction had slightly change which means the body was stable in the kick movement; the peak velocity of COM was found from the subjects jump up to the ready kick movement. In conclusion, 360° jump back kick needed highly demand for the body stabilization in order to allow the board to be broken.
KEY WORDS: 360° jump back kick, Power breaking.

INTRODUCTION:

Taekwondo is not only competition in the marches but also pursuit of personal skills. Breaking a one inch board is utilized for testing personal kicking motions and the ability of force control. The 360° jump back kick is more complex motion included rotation, jump and kick. Therefore, this motion has listed in the level promotion item. Also each one must break broad in Taiwan. Lee and Huang (2006) indicated regarding the kicking force, the 360° jump back kick is significantly smaller than the back kick and the jump back kick. The purpose of the study was to investigate the kinematic features of the 360° jump back kick in order to break a board.

METHOD:

Eight Taekwondo coaches (Height: 175.7±5.7cm; Weight: 75.0±7.9kg; Age: 28.0±4.2years) performed a kick movement five times to break a one inch board. Before each coach broke a board, the boney landmarks were attached. The segment coordination was defined by ISBS recommendation (Wu, 2003; Wu, 2005). The data of each mark was collected by ten Vicon cameras (MS13+, 250Hz) and all variables, linear and angular velocity of hip, knee and ankle in attack leg, displacement and velocity of center of mass (COM), were calculated by a custom written program of MatLab. The body segment parameters were adopted by Winter (1990). Standard deviation was evaluated the stabilization in displacement of COM. The statistic method was run by trials.

The 360° jump back kick was divided into six phases, prepare, step, rotation, kicking and contact phase (Figure 1). Prepare phase is defined as from ready position to the forward leg leaving the ground. Step phase is defined as the forward leg from leaving the ground to touch down on the ground. Rotation phase is defined as the forward leg being the support leg and whole body rotating toward the target until the support leg jumping from the ground. After the support leg jumping, it becomes the attack leg. Kicking phase is defined as the process of the attack leg breaking a board in the air. Contact phase is defined as the attack leg bending to minimal angle in knee, which is ready kick, and then knee extension to break the board. Thus, contact phase is a part of the kicking phase.

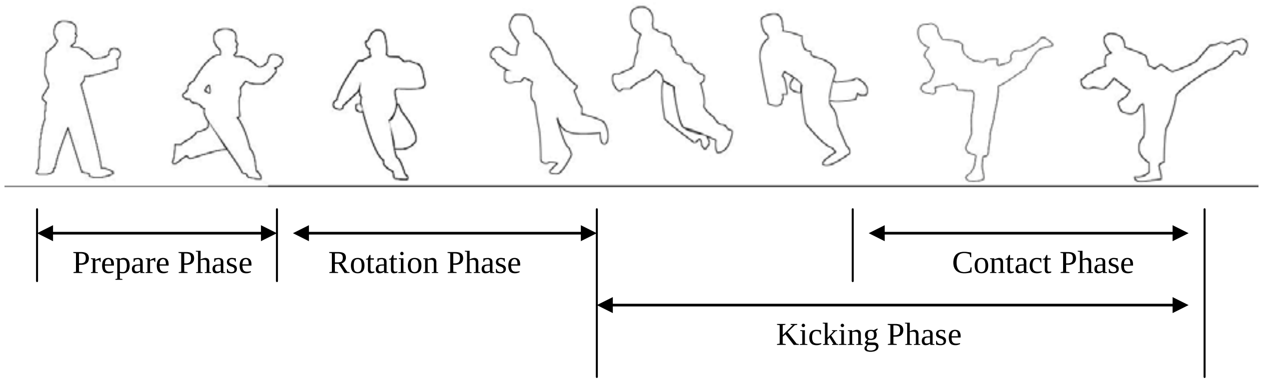

Figure 1. Illustration of 360°Jumping Back Kick

RESULTS:

Regarding linear and angular velocity of hip, knee and ankle in the attack leg, the peak velocity appeared timing (Table 1) which was not following the kinetic chain. For angular velocity, the sequence of each joint was knee, ankle and hip. For linear velocity, the sequence of peak value was hip, ankle and knee.

Table 1. Sequence of peak velocity value in the attack leg n=40

	Hip Joint		Knee Joint		Ankle Joint		Entire Time (sec)	
	M	SD	M	SD	M	SD	M	SD
Angular Velocity (deg/s)	759.75	651.30	587.40	539.48	714.98	715.53		
Appearing Timing (%)	98.70	3.97	96.30	6.44	98.57	4.42	0.91	0.14
Linear Velocity (m/s)	3.56	0.64	7.48	1.77	9.42	3.31		
Appearing Timing (%)	90.03	10.22	100	0.00	99.05	2.93		

In terms of stabilization during the whole movement, the standard deviation of COM displacement in two directions showed the subject's stabilization in each phase (Table 2).

Table 2. Standard deviation of COM in each phase n=40

	Prepare		Step		Rotation		Kicking	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Lateral Direction	0.01	0.01	0.01	0.01	0.07	0.04	0.16	0.11
Vertical Direction	0.01	0.01	0.02	0.01	0.17	0.15	0.64	0.17

The displacement of COM (Figure 2) illustrated the stabilization duration time between lateral and vertical direction. Turing to the compound velocity of COM (Figure 3), the peak velocity (Table 3) was found in the jumping movement to ready kick movement.

Table 3 Peak Compound velocity of COM and appearing timing n=40

	Compound Velocity of COM	
	Mean	SD
Peak velocity of COM (m/s)	6.07	0.91
Appearing Timing (%)	80.60	7.34
Entire time(sec)	0.91	0.14

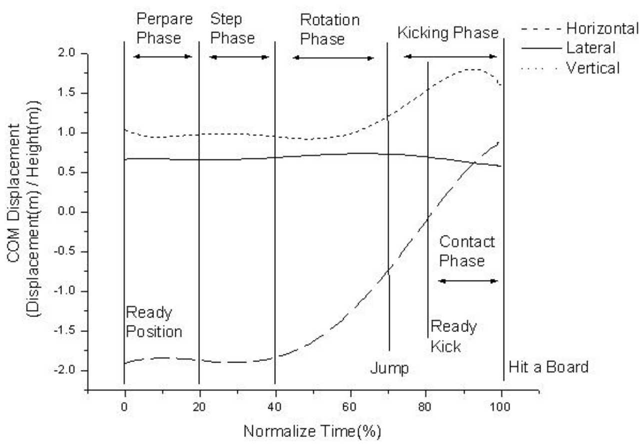

Figure 2. Displacement of COM

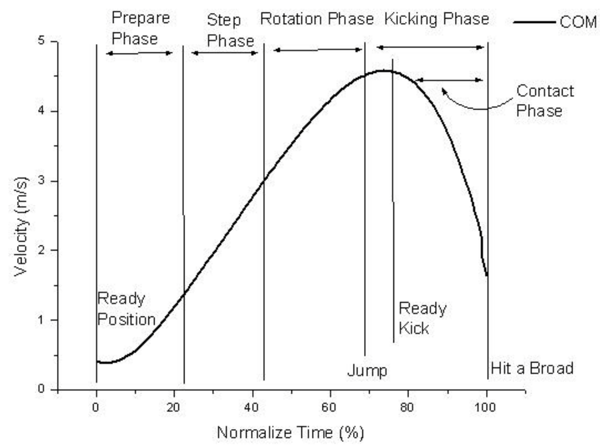

Figure 3. Velocity of COM

DISCUSSION: The kinetic chain of attack in the back kick is not followed and the sequence is hip, ankle and knee (Hong, Lina & Liu, 2006). Moreover, back kick, jump back kick and 360° jump back kick have the same pattern in the attack leg (Lee & Huang, 2006). However, in those two studies the target was a kicking bag rather than a board. Therefore, it can be seen the peak value of each joint's velocity is nearly kick a board because a board breaking is a main purpose resulting highly joint velocity to appear approximately at the end. Each joint can keep extension for breaking a board.

According to the 360° jump back kick is complex movement, the center of mass of body needs to keep stable under the dynamic kicking. The data shows that the lateral direction changes slightly from prepare phase to kicking phase. It can be explain the movement of COM is linear moving. The vertical direction also change slightly from prepare to rotation phase because of jumping after rotation phase. When whole body leaves the ground, the moving trajectory is controlled by the gravity. Thus, the lateral direction has to keep stability which allows body moving direction to contribute in the forward direction. Furthermore, during the kicking phase, the vertical direction is acted on by force of gravity and the moment of inertia is kept forward, so that when lateral is altered in a minimal range, the attack leg can aim the board and the attack force can focus on the feet in order to enhance a chance to break the target.

With regard to the COM velocity, the peak velocity has to appear before the attack foot hits the target (Houg, Lina & Liu, 2006; Lee & Huang, 2006). The 360° jump back kick is the same. The peak COM velocity is almost arrived in contact phase. Most of cases appear in the ready kick position, which is knee in minimized angle. From the point of the projectile movement, the attack leg needs enough time to extend the joints to break a board or target. The peak COM velocity must happen before hitting the target; otherwise, the attack leg cannot break the board.

CONCLUSION: In conclusion, the attack leg does not follow the kinetic chain in the 360° jump back kick. Using the 360° jump back kick breaks a broad needing the body stabilization. Both the lateral and vertical displacement had little change from prepare phase to kicking phase and from prepare phase to rotation phase respectively. Showing the stability of the body moving forward is one of important factor in the 360° jump back kick. Most of the peak velocity of COM was found from the jumping up to ready kick movement.

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DESCRIPTIVE ANALYSIS OF sEMG DURING THE WINDMILL SOFTBALL PITCH: PRE-FATIGUE AND FATIGUED

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The purpose of this study was to describe the upper and lower extremity muscle activation patterns of the windmill softball pitch during pre-fatigue and fatigued states. Biomechanical data concerning softball pitching is limited; however data examining muscle activations during a pre-fatigued and fatigued state during softball pitching are nonexistent. Just as there is prevalence to injury when performing the baseball pitch, that element of injury also arises in softball pitching. Thus in attempt to prevent injury, the muscle activations occurring during the pitching motion must be understood. Ten softball pitchers volunteered for the study. Participants were analyzed with surface electromyography, and motion analysis software. The muscle firing patterns were described during the phases of the softball pitch.

KEYWORDS: electromyography, muscle activation, overuse injury, shoulder injury, softball throwing, fastball

INTRODUCTION: Pitching injuries primarily encompass the shoulder, specifically the rotator cuff,and are typically caused by excessive pitching and improper mechanics (Loosli et al., 1992). A study examining the 1989 College Softball World Series revealed that windmill softball pitchers had a significant number of time-loss injuries (Loosli et al., 1992). It was also concluded that the injuries were directly related to pitching. There are no mandated recommendations on the number of pitches, innings pitched or rest between pitching outings for softball pitchers. It is also known that softball teams do not carry as many pitchers as baseball teams and softball pitchers could potentially throw as many as 2000 pitches within a three day tournament. In attempt to reduce the number of time-loss injuries sustained by windmill softball pitchers, it is important to understand the muscle activations during the pitching motion. Typically, injuries to the upper extremity that occur from softball pitching are overuse in nature (Pollack et al., 2005; Rojas et al., 2009). In attempt to understand the underlying mechanism of overuse injuries in the windmill softball pitch, it is not only important to understand the muscle activations during the pitching motion but also muscle activations during a fatigued state. Fatigue or overuse often leads to injury in pitchers. To date there are no know data examining muscle activations during a pre-fatigued and fatigued state of softball pitching. Therefore, it was the purpose of this study to examine upper extremity muscle activations during softball pitching in pre-fatigue and fatigued states.

METHODS: Ten softball pitchers (17.4±3.27years, 178.2±7.2cm, and 76.9±12.2 kg) volunteered to participate. All participants had recently finished their competitive spring high school softball seasons, and were deemed appropriately conditioned for participation. Data collection sessions were conducted indoors at the University's Health, Physical Education, and Recreation building and were designed to best simulate a competitive setting. All testing protocols used in the study were approved by the University's Review Board.

Participants reported for testing prior to engaging in resistance training or any vigorous activity that day. Palpation revealed the locations of the following muscles: bilateral gluteus maximus, bilateral gluteus medius, throwing arm biceps, triceps, deltoid and scapular stabilizers. Adhesive 3M Red-Dot bipolar surface electrodes (3M, St. Paul, MN) were attached over the selected muscle bellies and positioned parallel to muscle fibers using techniques described by Basmajian and Deluca (1985). Once all electrodes had been secured, manual muscle tests (MMT) were conducted for each muscle using techniques described by Kendall et al.(1993). The MMTs were used to identify the maximum voluntary isometric contraction (MVIC) for each muscle in attempt to establish baseline readings for each participant's MVIC to which all sEMG data were compared.

The MotionMonitor™ motion capture system (Innovative Sports Training Inc, Chicago IL) transmitted the surface electromyographic (sEMG) data via a Noraxon Myopac 1400L 8-channel amplifier. The signal was full wave rectified and smoothed based on the smoothing algorithms of root mean squared at windows of 100 ms, and sampled at a rate 1000 Hz. In addition, all sEMG data were notch filtered at frequencies of 59.5 Hz and 60.5 Hz respectively (Blackburn and Pauda, 2009).

Kinematic data were collected to event mark the phases of the pitching motion and these data were collected using The MotionMonitor™ motion capture system (Innovative Sports Training, Chicago IL). Prior to completing test trials, participants had ten electromagnetic sensors attached at the following locations: (1) the medial aspect of the torso at C7; (2) medial aspect of the pelvis at S1; (3) the distal/posterior aspect of the throwing humerus; (4) the distal/posterior aspect of the throwing forearm; (5) the distal/posterior aspect of the non-throwing humerus; (6) the distal/posterior aspect of the non-throwing forearm; (7) distal/posterior aspect of stride lower leg; (8) distal/posterior aspect of the upper stride leg; (9) distal/posterior aspect of non stride lower leg; and (10) distal/posterior aspect of non stride upper leg (Myers et al., 2005). Following the attachment of the electromagnetic sensors, an eleventh sensor was attached to a wooden stylus and used to digitize the palpated position of the bony landmarks.

Participants were allotted an unlimited time to perform their own specified pre-competition warm-up routine. Each participant threw a series of maximal effort fastballs for strikes toward a catcher located the regulation distance (12.2 m). The pitching surface was positioned so that the participant's stride foot would land on top of the 40 x 60 cm Bertec force plate (Bertec Corp, Columbus, Ohio) which was anchored into the floor. After five fastballs for strikes were thrown, the participants then threw a 2kg ball into a rebounder until they reported their maximum perceived fatigue. A scale of 0-3 (Kimura et al., 2007), with three being only able to make 15 more throws, was used to quantify fatigue. Once a fatigue of 3 was reported, participants completed 10 more throws with the 2kg ball before returning to the pitching surface to throw five maximum effort fastballs while in the fatigued state. Those data from the fastest pitch passing through the strike-zone for the pre-fatigue and fatigue deliveries were selected for analysis.

Data were analyzed in the current study using the statistical analysis package SPSS 15.0 for Windows. Data from the fastest strike for all sEMG and kinematic parameters were calculated for both pre-fatigue and fatigued states. Once measures of central tendency were calculated, a series of descriptive statistics were conducted. A MANOVA was run for each of the four phases comparing pre-fatigue to fatigue with an adjusted alpha level using Bonferroni correction to allow for multiple tests ($p \leq 0.01$). Phase 1, was defined as the start of pitching motion to the top of back swing (TOB). Phase 2, was from TOB to stride foot plant (SFP). Phase 3 was from SFP to ball release (BR), and Phase 4 was from BR to the completion of follow through.

RESULTS: There were no significant differences found between phases during pre-fatigue and fatigue conditions; $p \leq 0.01$. Means of muscle activations are graphically summarized in Figures 1-2.

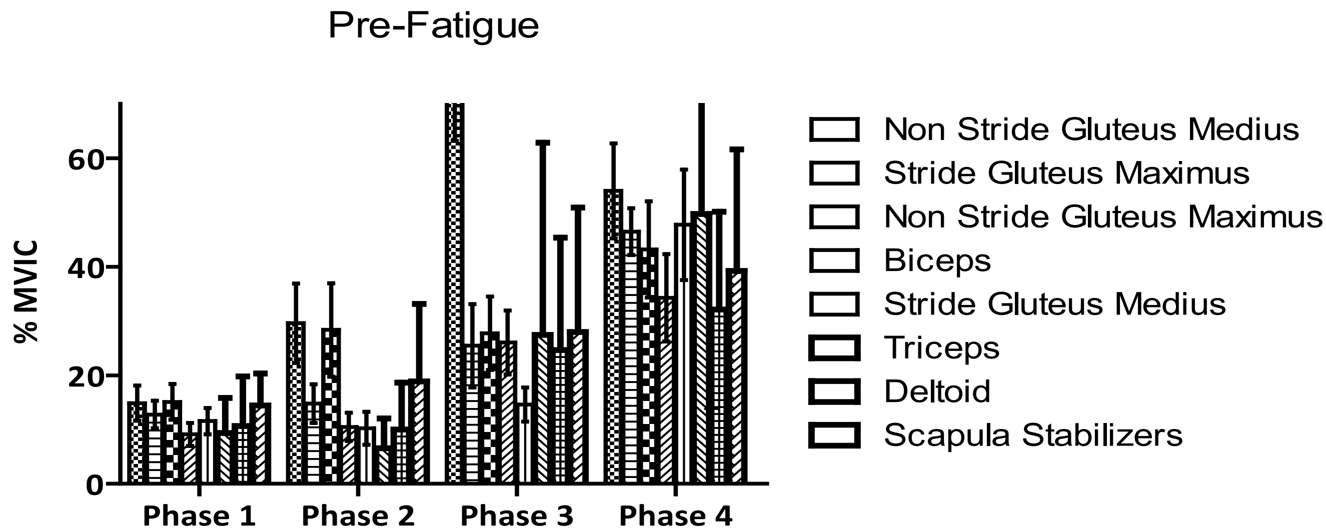

Figure 1. Muscle Activation as a %MVIC during the pre-fatigued state of the Windmill Softball Pitch.

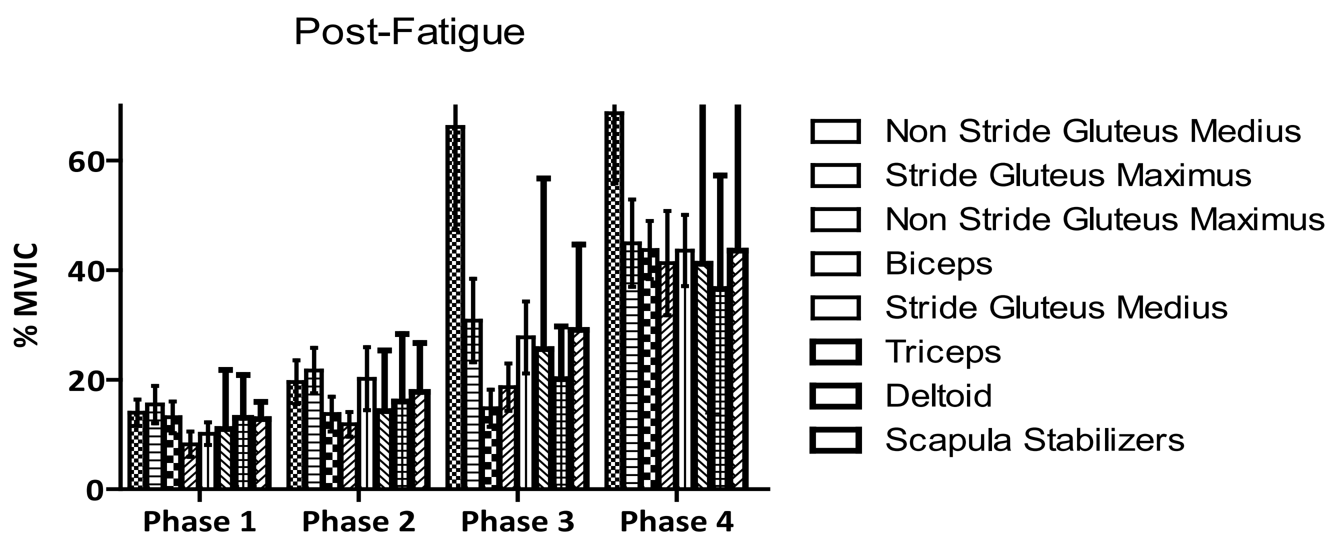

Figure 2. Muscle Activation as a %MVIC during the fatigued state of the Windmill Softball Pitch.

DISCUSSION: During the pre-fatigued Phase 1, the gluteal muscle group displayed the greatest muscle activation. The gluteal muscle group was activated during this phase in attempt to stabilize the pelvis for the stride into FC. Once fatigued the gluteal muscle group displayed similar activations while in the upper extremity, the triceps had greater activation than their pre-fatigue state.

Phase 2 revealed great activation of the non stride leg gluteus medius as well as increased overall gluteal activation. Again, Phase 2 is from TOB to FC which consists of continuing on single leg support to FC. Thus in attempt to keep the pelvis level, the gluteal group had increased activation while in single leg support. During the fatigued state the gluteals decreased in activation while the triceps increased; and biceps, deltoid and scapula stabilizers remained consistent.

Phase 3 displayed great activation of the gluteals, both stride leg and non stride leg, and greater activation of the upper extremity musculature. During this phase the windmill softball pitcher attempts to 'post' for ball delivery on the stride leg during stride foot plant, the non stride gluteal muscle group must hold the stride side hip upright, while the pitcher is balanced on the stride leg. Once fatigued, the gluteals decreased while the upper extremity musculature increased.
The follow through, or Phase 4, displayed an increase in all muscle activations excluding the non stride gluteus maximus in attempt to decelerate the entire motion. The same trend was shown in the fatigue trail.

CONCLUSIONS: The current study was able to quantify and describe muscle activations for the upper and lower extremity during the windmill softball pitch both pre-fatigue and in a fatigued state. As this is the only investigation of our knowledge examining upper and lower extremity muscle activations in the fatigue state, only generalizations for muscle activation patterns were presented. It is speculated that when comparing fatigue versus non fatigue conditions that practitioners would want similar muscle activations in attempt to have decreases in kinematic alterations. More studies need to be conducted in attempt to validate our results with a higher level of evidence of muscle activation and movement kinematics during dynamic muscle fatigue. In addition more studies need to explore the relationship of decreased gluteal muscle activation and increased upper extremity activation in the fatigued state and their implications to injury.

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