AN ABSTRACT OF THE DISSERTATION OF

Grace M. Golden for the degree of Doctor of Philosophy in Nutrition and Exercise Sciences presented on May 3, 2007.
Title: Investigation of the Biomechanics of Running and Rapid Change-of-Direction Tasks

Abstract approved:

Mark A. Hoffman, Michael J. Pavol

Rapid change-of-direction tasks have been associated with non-contact anterior cruciate ligament (ACL) injuries and females are more likely to suffer this injury compared to males. The purpose of this investigation was to evaluate the biomechanical behavior of running and three rapid change-of-direction tasks and determine whether sex differences exist across tasks. The biomechanical factors included knee and hip kinematics and kinetics, in addition to static lower extremity alignment and dynamic hip strength. Data were collected in a controlled laboratory setting on healthy collegiate female and male basketball and soccer players (N = 21). Three-dimensional kinematics and kinetics were recorded in conditions of running, lateral false step, sidestep cut, and a combination of a lateral false step with a sidestep cut. Static lower extremity alignment was represented by measurement of standing Q-angle. Hip strength was determined by measuring isokinetic eccentric hip abduction strength. Three queries were addressed: 1.) sex differences in the kinetics and kinematics of the knee and hip during running and rapid change-of-direction tasks, 2.) the effect of sex on relationships of standing Q-angle and hip strength to frontal plane knee biomechanics, and 3.) description of the patterning of knee and hip biomechanics across tasks and sex. The results of this study suggest knee abduction and internal moments of knee adduction and hip abduction increase when athletes step laterally in combination with a sidestep cut compared to all other tasks. Females demonstrate differences in hip position and loading compared to males across the four tasks. The data did not support evidence of an effect of sex on standing Q-angle or hip abduction strength. Additionally, no relationships between sex, standing Q-angle, or hip abduction strength to frontal plane knee biomechanics were found. Four patterns of knee and hip kinematics and kinetics were found to describe lower extremity biomechancis during running and rapid change-of direction tasks. In summary, the addition of frontal plane motion and loading during change-of-direction tasks significantly affected

knee position in a manner associated with ACL injury. Differences in lower extremity kinematics and kinetics between females and males appear to be specific to hip position and loading.

Investigation of the Biomechanics of Running and Rapid Change-of-Direction Tasks

by
Grace M. Golden

A DISSERTATION

submitted to

Oregon State University

in partial fulfillment of
the requirements for the
degree of

Doctor of Philosophy

Presented May 3, 2007
Commencement June 2007

Doctor of Philosophy dissertation of Grace M. Golden
Presented on May 3, 2007

APPROVED:

Co-Major Professor, representing Exercise and Sport Science

Co-Major Professor, representing Exercise and Sport Science

Chair of the Department of Nutrition and Exercise Sciences

Dean of the Graduate School

I understand that my dissertation will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my dissertation to any reader upon request.

Grace M. Golden, Author

ACKNOWLEDGEMENTS

The author expresses sincere appreciation to: Mark A. Hoffman, Michael J. Pavol, Louis R. Osternig, Rod A. Harter, for their enduring guidance, mentoring, and patience through my adventures in further education and during this entire process of trying to find answers to some of life's interesting questions.

My best cheers to Kim Hannigan-Downs, Sam Johnson, and Kristin Lage for their wonderful friendships- your input, insight, and motivation inspires me to be a perpetual student of life and its lessons…….

Brian and Amy and the boys…..my time in Corvallis allowed me to find very good friends- perhaps one of the best gifts of all through this entire process.

Lastly, thanks and love to my family and friends for their unconditional support throughout my life, teaching me to enjoy all the adventures along the way.

CONTRIBUTION OF AUTHORS

Dr. Mark A. Hoffman assisted with study design, statistical analysis, interpretation, and writing of the manuscripts. Dr. Michael J. Pavol assisted with study design, biomechanical program design and analysis, data collection, interpretation and writing of the manuscripts.

TABLE OF CONTENTS

TABLE OF CONTENTS (Continued)

LIST OF FIGURES

LIST OF TABLES

GENERAL INTRODUCTION

Anterior cruciate ligament (ACL) injuries continue to have a significant impact on competitive athletes. Athletes participating in sports requiring jump landings and rapid stopping or changes in direction are more susceptible to non-contact ACL rupture than a contact related injury.[3, 15, 26, 45, 46, 97] These behaviors are common in the sports of basketball and soccer and injury statistics provide conclusive evidence that females are more likely to suffer an ACL injury than males.[1, 5] For these reasons, researchers are motivated to understand how the ACL is injured and why there is a sex bias.

The at-risk knee position for ACL injury is one of internal rotation and abduction when the knee is near full extension.[15] This clinically observed knee position has also been shown to increase loading on the ACL during in situ studies of the cadaver model.[84, 85, 125] One primary area of research of at-risk maneuvers believed to cause this dangerous knee position is the investigation of sidestep cutting. Sidestep cutting represents a powerful plant and cut maneuver in which the athlete abruptly changes direction 30-90 degrees from the original path of travel. Although controlled biomechanical studies are designed to measure the biomechanical behavior of the lower extremity without inducing ACL injury, investigators are still able to determine if knee positions and loading factors similar to the proposed mechanism of injury are possible during sidestep cutting tasks. There is support for arguments of sagittal and frontal plane loading as being culpable in the ACL injury mechanism during sidestep cutting tasks.[28] It seems prudent to consider that a relationship exists between the two loading mechanisms, in addition to a possible role of transverse plane loading.

Due to the complex nature of ACL injury, a multi-factorial approach to quantifying precipitating factors to the ACL mechanism of injury and differences between females and males has been adopted.[28, 86, 87] This approach considers anatomical, hormonal, neuromuscular, and biomechanical issues. This study was designed to determine if there are direct relationships between anatomical and neuromuscular factors and measurements of biomechanical behavior during running and rapid change-of-direction tasks. At the same time, control of potential contributing hormonal influences was included in the study design.

Typically, anatomical factors are considered non-modifiable and represent characteristics such as ACL structure (volume, cross-sectional area, mass, loading behavior)[20, 67, 130], lower extremity alignment (Q angle, foot pronation)[18, 77], and joint configuration.[2, 21] Q-angle is known to differ between males and females[59] and the general belief is individuals with a larger Q-angle may be more susceptible to ACL injury.[60, 94, 100] There is scant evidence that athletes with a larger Q-angle will also display greater knee abduction position during dynamic tasks. In one study, researchers did not find a relationship between Q-angle and knee abduction during a single leg squat.[105] Recent studies have reported Q angle is affected by quadriceps muscle strength and potentially changeable. This leads to the possibility of Q-angle also being affected by muscular strength of the hip musculature, i.e. the hip abductors, which affect femur position due to their anatomical attachments. Although knee abduction has been found to be related to strength of the hip abductors during a jump landing task[64], there has not been an investigation of the relationship between hip abduction strength and frontal plane position of the knee during sidestep cutting tasks. One focus of this study was to determine if hip abduction strength and a standing measure of Q-angle were related to frontal plane knee and hip position during change of direction tasks.

Hormonal influences are directly related to the presence of systemic concentrations of estrogen, progesterone, and testosterone. Estrogen is believed to affect the mechanical properties of the ACL ligament in addition to reducing ACL cellular metabolism.[75, 76, 132, 133] Females experience greater fluctuations in estrogen and progesterone concentrations compared to males due to the cyclical effects of the menstrual cycle. Estrogen and progesterone concentrations are the lowest during menses and reach peak levels near ovulation. At this time, is not clear if ACL laxity changes as a direct result of changes in hormone concentrations or if females are more susceptible to injury at different times across the menstrual cycle.[11, 50, 116] In order to control for the potential effect of the menstrual cycle and variability in hormone concentrations for females, the period of time in which data was collected in this study was standardized to be during the follicular phase for female subjects.

Neuromuscular factors include muscular strength and recruitment patterns during dynamic tasks. Coordination of muscular actions enhances joint stability and position during these tasks. With regards to ACL loading and knee position, the primary muscles of interest have been the quadriceps and hamstrings. The quadriceps muscle is believed to be antagonist to the ACL, meaning forceful contraction will increase strain on the ACL, whereas the hamstrings are protective of the ACL. Consideration of muscles indirectly affecting the knee is also warranted. Proximal muscles at the hip, while directly influencing hip motion, may secondarily affect knee position due to their contribution to controlling the position of the femur in the frontal, sagittal, and transverse planes. Specifically, the hip abductors contribute to frontal plane and transverse plane motion at the hip, causing femur abduction and rotation.[31] To date, the effect of sex on neuromuscular factors is unclear. Although females have been reported to have a quadriceps dominant recruitment pattern during jump landings and sidestep cutting tasks by some researchers [52, 55, 82], others have not found the same evidence of differences in thigh muscle activation patterns between males and females.[34] What is not known is whether activation patterns and overall strength of hip musculature have a direct effect on knee position and loading patterns during athletic maneuvers. One objective of this study was to determine if there are differences in hip abduction strength between females and males, and if athletes with less hip abduction strength demonstrate greater frontal and transverse plane angles and loading at the knee.

Biomechanical behavior of females and males during rapid change of direction tasks has not been found to conclusively differ. Knee abduction has been reported to be a predictor of ACL injury in females during jump landing tasks[54], yet the role of knee abduction as a predictor during sidestep cutting tasks is less clear. Although differences between females and males in frontal plane hip and knee position have been found during running[35], the evidence is less clear during sidestep cutting tasks. There is lack of agreement as to whether knee kinetics and kinematics differ between males and females during sidestep cutting tasks.[92, 95, 107, 119] Currently, investigators are considering what influence hip position and loading has on knee position during these tasks in order to better understand the ACL injury mechanism. One study found females exhibit less peak hip abduction compared to males during sidestep cutting tasks.[107] It is possible one reason investigators have not been able to identify how the sidestep cutting maneuver is directly associated with ACL injury is due to constraints adopted during study design. One contributing factor worthy of consideration is whether directing the plant leg during a sidestep cutting maneuver in a direction opposite of the intended change of direction results in a more dangerous knee position than sidestep cutting alone. Stepping laterally with a single step within a running stride has been termed a lateral false step[43] and conceivably the lateral false step can be performed as a rapid change-of-direction task by itself or in combination with a sidestep cutting task. To date, there has not been a direct comparison of sidestep cutting to a combination of a sidestep cut and a lateral false step. To evaluate these potential differences, this study was designed to directly compare knee and hip position and loading patterns during sidestep cutting and a combination of the two techniques. Additionally, comparisons of lower extremity kinematic and kinetic behavior across normal running and a lateral false step alone were included in the study design. This investigation was designed to not only compare peak joint angles and moments of the knee and hip in the frontal and transverse planes, but also determine if patterns of lower extremity position and loading are different between the tasks and if females and males exhibit differences in these patterns.

This study was designed to measure anatomical and biomechanical factors related to ACL injury, at the same time, controlling for potential hormonal influences. A 3-dimensional motion analysis system was used to measure lower extremity kinematics and kinetics during running and three rapid change of direction tasks: 1.) lateral false step, 2.) sidestep cut, and 3.) combination of the lateral false step and sidestep cut). Additionally, measurement of static lower extremity alignment (standing Q-angle) and peak eccentric hip abduction strength was performed. Potential hormonal influences on biomechanical and strength behaviors were controlled for females by collecting data during a specific phase of the menstrual cycle.

The purpose of this study was several-fold. First, comparison of standing Q-angle, hip abduction strength and position of peak torque production during eccentric hip abduction in collegiate female and males athletes was performed. The hypothesis was that differences in standing Q-angle and hip strength (normalized to body height and body weight) would be found between males and females. Secondly, determination of relationships to standing Q-angle and hip abduction strength between knee and hip position and loading patterns during running and rapid change of direction tasks was conducted. The expectation was that athletes demonstrating greater standing Q-angles and weaker hip abduction strength would exhibit greater frontal plane knee and hip peak positions and loading during running and change of direction tasks. The next objective was to compare peak knee and hip kinematic and kinetic measures between conditions of running, a lateral false step maneuver, sidestep cutting and a combination of a sidestep cut with a lateral false step. The central hypothesis was that a combination of the lateral false step with the sidestep cutting maneuver would result in greater frontal and transverse plane knee and hip loading and positions which are believed to increase loading of the ACL in a manner consistent with the proposed mechanism of injury. Additionally, females were expected to demonstrate different lower extremity alignment and loading patterns compared to males. The expectation was that outcomes of this study would help to better understand the mechanism of ACL injury during rapid change of direction tasks and why females are more susceptible to this injury.

Sex differences in kinematics and kinetics of the knee and hip during running and rapid change-of-direction tasks

Grace M. Golden, Michael J. Pavol, Mark A. Hoffman

INTRODUCTION

The disproportionate incidence of non-contact anterior cruciate ligament (ACL) injuries between males and females continues to be an enigma. Non-contact injury mechanisms account for 70-78% of all ACL ruptures in the athletic population[4, 25], with female soccer and basketball players 2-4 times more likely to suffer ACL injury than males in matched sports.[1, 3] Non-contact ACL injury is most often associated with a sudden deceleration, such as occurs with abrupt stopping[12, 25], double- or single-leg jump landing[12, 25], or a rapid change of direction[2, 4, 5], i.e. cutting. These injuries are generally believed to occur near foot contact when the knee is relatively extended, internally rotated, and abducted.[15] This is consistent with the in situ findings that the ACL is loaded when the knee is relatively extended (0-30º) and subjected to internal rotation and abduction external torques.[18, 19, 32] Therefore, researchers have attempted to determine whether these combinations of knee positions and loadings occur during high risk athletic maneuvers and whether females exhibit them to greater extent than males.

It remains unclear whether sex differences in knee kinematics and kinetics exist for rapid change-of-direction tasks, such as sidestep cutting (SSC). Investigators have determined females demonstrate greater knee abduction[17, 21] and less internal rotation[22] and knee flexion[17, 22] compared to males in SSC tasks. However, equally persuasive investigations have not established a sex difference in knee kinematics for SSC tasks.[26, 30] Intuitively, sex differences in knee loading patterns should exist in the presence of differences in knee kinematics. Researchers have reported such a relationship, with females experiencing greater external knee moments in abduction[20, 21] and smaller external knee moments in internal rotation[20] compared to males during SSC tasks. Similarly, Sigward and Powers[30] found females demonstrate greater peak external moments in abduction compared to males, despite no differences in knee kinematics. Conversely, Pollard et al.[26] did not detect any sex differences in knee kinetics or kinematics during SSC tasks. The lack of consensus on sex differences in knee position and loading between females and males may be attributable to subtle differences in how the SSC tasks are conducted across studies.

Recently, investigators have considered what role hip kinematics and kinetics play in contributing to knee position and loading patterns with regard to ACL injury mechanisms during sidestep cutting. McLean et al.[22] suggest that increased hip internal rotation in addition to increased knee abduction is related to the amount of knee adduction loading (external knee abduction moments). While these researchers found females to display greater hip internal rotation and knee abduction in the presence of greater knee adduction moments compared to males during sidestep cutting, these sex differences are not consistent across studies. Pollard et al. [27] found females exhibit less peak hip abduction than males without an effect of sex on knee kinematics and kinetics or internal hip rotation. Based upon the conflicting results of these studies, it is not clear if transverse or frontal plane hip kinematics contribute to knee adduction loading. Therefore, further investigation is warranted.

Knee abduction has been advocated as a dominant risk factor for ACL injury for females in jump landing and cutting maneuvers.[14, 22] However, the relationship between knee rotation and ACL injury in vivo remains unclear with respect to SSC tasks. While differences in knee rotation between females and males have been found, peak values are within the passive range of motion limits of the knee.[22] Similarly, the relative contribution of sagittal plane loading on the ACL remains controversial. Females have been found to achieve maximum knee flexion earlier than males in SSC tasks[24] suggesting males may spend more time controlling loading through the sagittal plane as compared to females. However, it is not clear whether females perform dynamic tasks with less knee flexion. Malinzak et al.[17] determined females demonstrate less knee flexion than males during running and SSC tasks. McLean et al.[24] found overall flexion-extension range of motion to be similar between males and females and in a follow-up study McLean et al.[20] concluded sagittal plane loading during a sidestep cutting task was insufficient to load the ACL in isolation but may indirectly contribute to the injury mechanism due to its potential influence on frontal plane knee kinematics and kinetics. This is in agreement with researchers who reported a combination of anterior tibial force and external valgus moment to be dangerous in a relatively extended knee.[18] DeMorat et al.[8] investigated simulated in vivo quadriceps contractions at the knee and concluded that aggressive quadriceps loading (4500N) when the knee is in slight flexion (20º) produces significant anterior tibial translation and ACL injury. This supports the hypothesis of sagittal plane loading with intrinsic muscular forces playing an important role in ACL strain. The major limitation to this finding is whether athletes can generate such large quadriceps forces at relatively low knee flexion angles. In light of these differing premises, it seems most appropriate to consider a combination of sagittal, frontal, and transverse positions and loading behavior as culpable in ACL injury.

It is plausible the SSC task, as currently defined and investigated, is not entirely representative of the high risk movement associated with the change-of-direction tasks exhibited when the ACL is reported to be injured, and this may in part explain discrepancies in study outcomes. Other variations of the SSC may provide a more ecological representation of the maneuvers used to alter running or change direction by athletes and therefore be sensitive enough to elicit differences in lower extremity kinematics and kinetics which may help to explain how the ACL is injured when changing directions. Typically, the SSC task is investigated as a pure change of direction occurring at 35-55 degrees from the original path of travel.[10, 17, 20, 21, 26, 30] Specifically, the athlete performs a run-to-cut maneuver in which straight ahead running is interrupted by a powerful acceleration into a different direction by the stance limb. However, athletes routinely display movement behaviors which are similarly patterned after this maneuver, but may be different enough to affect knee position and loading during the redirection. One such variation of the SSC is a lateral false step (LFS). A LFS is performed by stepping laterally with one step during a running stride, and then continuing to run forward in the original direction of travel. The length of this LFS may differ across athletes. Previously, the LFS of 20 and 35% of body height has been found to significantly affect knee kinematics and kinetics when compared to running.[11] Similarly, the LFS can be performed in combination with a SSC (LFS-SSC). In this case, the SSC is initiated by stepping laterally, away from the final intended change of direction, as part of the cutting maneuver. The LFS-SSC includes a false step into a direction in which the athlete does not intend to continue to run and may serve to increase the deception associated with the change of direction in order to gain advantage over the opponent. Constraining the rapid change-of-direction maneuver to a pure SSC may ignore different strategies athletes use to redirect running in agility dependent sports.

Since athletes utilize different movement techniques to rapidly change direction while running, investigation of variations in sidestep cutting may be helpful in determining how these task differences affect knee and hip position and loading patterns. Specifically, investigation of normal running (RUN), a lateral false step maneuver (LFS), sidestep cutting (SSC), and a combination of the LFS and SSC can serve to aid our understanding of which movement patterns are potentially more dangerous for the ACL. The LFS represents a laterally directed step within a running stride, followed immediately by a step back to the original direction of travel and a continuation of running straight ahead in the original direction of travel.

The combination of the LFS and SSC is characterized by a SSC in which the stance limb first performs a LFS to the side opposite the direction of the cut. It is not known: 1.) whether the LFS maneuver differs from the SSC task with regards to peak knee and hip angles and moments, 2.) whether the combination of a LFS and SSC results in more deleterious knee and hip loading and positions than either maneuver alone, or 3.) whether sex differences in knee and hip kinematics and kinetics exist between these tasks.

This study was designed to systematically investigate variations of a SSC maneuver and further clarify the presence of sex differences in knee and hip kinematics and kinetics and the potential for differences in ACL loading between the tasks. Therefore, the purpose of this study was to compare the 3-dimensional hip and knee kinematics and joint moments in collegiate athletes during conditions of normal running and three change-of-direction tasks. The first hypothesis was that female basketball and soccer athletes would exhibit greater knee abduction and internal rotation, and greater hip adduction and internal rotation compared to males from matched sports across all tasks. Secondly, it was hypothesized that females would exhibit greater internal joint moments of knee adduction and external rotation and hip abduction and external rotation compared to males across all tasks. Lastly, it was theorized that significant differences in knee and hip kinematics and kinetics would exist across running and change-of-direction tasks. All comparisons were constrained to a period in stance phase from initial contact to 30º of knee flexion.

METHODS

Twenty-one subjects (11 females: 19.8 ± 1.5 yrs, 175.8 ± 6.3 cm, 69.9 ± 6.6 kg; 10 males: 20.7 ± 2.5 yrs, 188.6 ± 15.9 cm, 89.8 ± 21.7 kg) , basketball and soccer athletes from four college and university programs, were enrolled in the study. During initial screening, subjects completed demographic and health history questionnaires. Exclusion criteria included a past history of any reconstructive surgical procedure for the treatment of a lower extremity injury within the past twelve month period. All subjects were required to have an intact ACL. They could not be currently receiving supervised medical care for an injury to the upper or lower extremity preventing their participation in regular conditioning or practice sessions nor could they have suffered an injury to the lower extremity resulting in time loss from participation exceeding three weeks within the past six months. Additionally, female subjects were screened for a regular menstrual cycle (28-32 days[28, 29]) for the past 3 months and participated in the testing session within 1-9 days of the start of menses, i.e. the follicular phase (6.5 ± 1.8 days). All participants were of equivalent sport experience (females: 8.45 ± 1.97 yrs, males: 7.00 ± 1.76 yrs; $p > 0.05$) and participating in either off-season conditioning (basketball) or a non-traditional competitive season (soccer). Recruitment and participation of subjects was approved by a University Institutional Review Board for the Protection of Human Subjects and all subjects provided informed consent prior to participation.

Kinematic and kinetic data acquisition were performed with a Vicon 612 optical motion capture system utilizing 9 high-resolution MCam2 digital cameras (Vicon, Lake Forest, CA), interfaced with two Bertec Model 4060-08 force plates (Bertec, Columbus, OH) that were configured to function as a single force plate. Sixteen markers (9mm diameter) were affixed bilaterally on the lower extremity with double sided tape in order to identify body segment positions during static and dynamic trails. A customized marker set, based upon the work of Vaughan et al.[31] was utilized for this study (5th metatarsal, heel, ankle, tibia, knee, thigh, anterior superior iliac spine, and posterior superior iliac spine). Kinematic data were collected at 120Hz and kinetic data were collected throughout foot contact at 1080 Hz.

Figure 1 illustrates the laboratory set up for the running and change-of-direction tasks. In order to identify the path of travel for RUN and SSC conditions, yellow tape was affixed to the floor in a line crossing the center of the force plate. A second, parallel, blue tape line was placed on the floor, crossing the left corner of the force plate to indicate the path of travel for the LFS and LFS-SSC conditions. A divergent tape line, originating from the force plates, was placed at an angle of 45º to the left of the other lines to identify the pathway for subjects to successfully perform the SSC and LFS-SSC change-of-direction maneuvers. Colored adhesive tape was applied on the force plate to indicate a distance of 30% of body height for LFS and LFS-SSC conditions. This distance was measured perpendicularly to the right from the center point of the tape line crossing the left corner of the force plate. Cameras were positioned so each marker was visible by multiple cameras when the right foot was in contact with the force plate. Each trial consisted of a running a distance of approximately 10m from start to finish.

On the day of testing, subjects reported to the testing laboratory. Height (cm) was measured (with shoes on) with a wall mounted stadiometer. Weight (lbs) was recorded with a floor scale and later converted to kg. Next, subjects performed a 5 minute warm-up by running on a treadmill (2.5-3.1 $m \cdot s^{-1}$). During this time, adhesive tape was applied to the force platform at a location indicating a distance equivalent to 30% of body height from the line of travel for the LFS and LFS-SSC conditions.

After the warm-up period, subjects were allowed 3-5 practice trials for each of the running and change-of-direction tasks in the order or RUN, LFS, SSC, LFS-SSC. These practice trials enabled researchers to determine the most effective starting location so that the running stride prior to force plate contact was fluid and consistent, and improved subject familiarization with the specific tasks. Throughout the practice and testing trials, the speed of the subject was monitored through the use of timing lights placed 1.5 m to the rear and front of the force plates, at a total distance of 3 m apart. Trials were considered acceptable when speed equated to 3.5 $m \cdot s^{-1}$ ± 10% (3.15 - 3.85 $m \cdot s^{-1}$). Recovery was accommodated by having subjects walk back to the starting location between each trial. Verbal instructions for running speed and technique were standardized for all subjects. Subjects were directed to perform all trials without a reduction of speed into the RUN, LFS, SSC, and LFS-SSC maneuvers, discouraging leaping or hesitating prior to force plate contact. Testing trials were discarded if the subject made foot contact on the force plates with the left foot, failed to make contact with the force plates with the entire right foot, missed the LFS and LFS-SSC tape mark on the force plate, initiated the LFS or LFS-SSC maneuver by crossing the left foot to the right of the directional guide placed on the floor, or completed the SSC or LFS-SSC cut outside a range of 45º ± 10º. Colored tape was placed on the floor to indicate the boundaries of 35-55º.

Following familiarization of running and change-of-direction tasks, markers were affixed to the subjects. Each subject wore form fitting spandex shorts of a dark color and a pair of their own team-issued running shoes for data acquisition trials. Foot markers were placed on the outside of the running shoe. A static trial was collected for calibration of segmental orientation, with the subject positioned in anatomical neutral (arms at 90º abduction) while standing on the force plates. Ten successful trials were recorded for each of the four conditions in the order of: RUN, LFS, SSC, and LFS-SSC. During each trial, kinematic and kinetic data were measured for the subject's right limb throughout contact with the force plate during stance phase. Subsequent to the completion of all running and change-of-direction trials, foot length (shoe length) and pelvis, knee, and ankle width were measured with a joint anthropometer (Lafayette Instrument Company, Lafayette, IN). Three measurements were obtained for each measurement by the same researcher and averaged for data analysis. All data were collected in a single testing session.

Three degrees of freedom were assigned to the major joints of interest, with the corresponding rotations of flexion-extension, abduction-adduction, and internal-external rotation for the hip, knee, and ankle joints. Raw data for all dynamic trials were used to determine the three-dimensional paths of the markers and cubic spline interpolation was performed to fill small gaps (less then 50ms) with Vicon Workstation software. Kinematic data were filtered using a fourth-order zero-lag Butterworth low-pass filter with a cut-off frequency of 16Hz, as determined through a residual analysis.[33] Marker positions from the static, standing trial and anthropometric measures were used to establish a set of transformations for determining joint center locations and body segment orientations. These transformations were used to determine the paths of the joint centers and the orientations of the body segments over the entire stance phase of running. From these, the joint angles (degrees) at the stance knee and hip were determined by the convention of Grood and Suntay.[13] Knee internal rotation angles during the static trial were defined to be zero. The computed kinematics were also combined with the force plate data to determine the internal joint moments acting at the knee and hip during stance using 3-dimensional inverse dynamics methods. Segmental inertial parameters were determined as outlined by de Leva for female and male data.[6] Joint moments were expressed with respect to the joint axes of rotation. Joint moments (Nmm) were normalized to body height (mm) and body weight (N) and reported as a percentage of body height (bh) and body weight (bw). A customized Matlab program (Student version 7, Mathworks, Natick, MA) was created to extract peak joint angles and moments from initial contact to 30º of knee flexion during stance for all dynamic trials. An average of the peak values from the ten trials of each condition by each subject was used for statistical analysis. Ensemble-average curves of the joint angles and moments were created and normalized to 100% of stance for each subject and condition, and these were then ensemble-averaged across subjects for each condition. In constructing the ensemble-average curves, cubic spline interpolation was used to determine the values at each % of stance.

Separate two-way repeated measures ANOVA's (2 x 4) were used to analyze the effect of sex and the four conditions on: 1.) peak knee angles of abduction and internal rotation, 2.) peak knee moments of extension, adduction, and external rotation, 3.) peak hip angles of adduction and internal rotation, and 4.) peak hip moments of abduction and external rotation. Post hoc comparisons between the four conditions were conducted utilizing the Tukey's HSD test. The level of significance was set at $\alpha = 0.05$. (SPSS software, version 13.0, Chicago, IL).

RESULTS

There were no significant interactions between sex and condition for any of the peak knee and hip angles and moments analyzed over the initial part of stance, i.e., with the knee flexed less than 30° ($p > 0.05$). The effects of condition thus did not differ between sexes, nor did the effects of sex differ between conditions.

The analysis revealed a main effect of condition on peak knee abduction, hip adduction, and hip internal rotation angles over the initial part of stance ($p < 0.001$; Table 1). Post hoc analysis showed all three change-of-direction tasks resulted in greater peak knee abduction and lesser peak hip adduction than RUN ($p < 0.05$). In addition, the combined LFS-SSC task produced greater peak knee abduction and lesser peak hip adduction than did either the LFS or SSC alone ($p < 0.05$). In contrast to RUN, the hip was abducted throughout the initial part of stance for each of the change-of-direction tasks. Both the SSC and LFS-SSC tasks involved greater peak hip internal rotation than RUN and the LFS-SSC task also exhibited greater peak hip internal rotation than LFS ($p < 0.05$). Ensemble-average curves for position data for the entire stance phase demonstrate the differences in knee abduction (Figure 2) and hip adduction and internal rotation (Figure 4) across conditions.

There was a significant main effect of condition on peak knee adduction, hip abduction, and hip external rotation moments over the initial part of stance ($p < 0.001$; Table 1). All three change-of-direction tasks involved greater peak knee adduction moments than RUN and the combined LFS-SSC task resulted in greater peak knee adduction moments than either LFS or SSC alone ($p < 0.05$). In addition, the LFS-SSC task involved lesser peak hip abduction and greater hip external rotation moments than RUN and LFS ($p < 0.05$). Ensemble-average curves across the entire stance phase for moment data demonstrate the differences in knee adduction (Figure 3) and hip adduction and external rotation moments (Figure 5) across tasks.

Sex differences were present for peak hip adduction angle ($p = 0.04$) and peak hip external rotation moment ($p = 0.02$) over the initial part of stance. Females demonstrated greater peak hip adduction angles (or lesser minimum hip abduction angles) and smaller hip external rotation moments than males (Table 2). Otherwise, there were no differences between males and females for any of the other peak angles and moments analyzed.

DISCUSSION

The results of this study support the premise that lower extremity biomechanics significantly differ between running and change-of-direction tasks. In addition, these data provide evidence of differences in kinematics and kinetics between alternative methods used to change direction. One primary movement associated with ACL injury when changing direction has been the sidestep cutting maneuver. Our results indicate variations of this maneuver may be associated with similar or more dangerous frontal plane joint alignment and loading when compared to sidestep cutting.

The peak knee abduction values observed during this study confirm sidestep cutting increases knee abduction compared to running. However, contrary to the work of Ferber et al.[9], we did not detect differences in knee abduction during running between females and males. Researchers suggest knee abduction is one of the precipitating factors of ACL injury during sidestep cutting.[21] While our sidestep cutting results differ from McLean et al.[24], who reported that females perform sidestep cutting tasks with more knee abduction than males, they were in agreement with the results of Sigward et al.[30] and Pollard et al.[26] Our data provide evidence that stepping laterally in combination with the sidestep cut will result in greater knee abduction than sidestep cutting alone. McLean et al.[23] have suggested the knee abduction associated with sidestep cutting may not be of sufficient magnitude to directly elicit a non-contact ACL injury. Our results suggest augmenting the frontal plane component of a sidestep cut with the addition of a lateral step increases knee abduction, possibly explaining how athletes are at risk of increased ACL loading when changing direction with this type of technique.

The lack of effect of task on knee internal rotation was consistent with previous investigations in which running, sidestep cutting, and cross-over cutting have been examined.[17] Based upon observation of our ensemble-average curves and peak angle data, the knee was internally rotated during early and mid-stance in all of the tasks. This is appears to be consistent with the known relationship between knee flexion and internal rotation as part of the reversal of the screw-home mechanism. It is also possible the observed internal rotation is also a product of the combination of increasing knee flexion and adduction during early stance. These results differ from those of McLean et al[23], who reported periods of absolute knee external rotation at foot contact during sidestep cutting. These inconsistencies between studies may be due to differences in how anatomical neutral was defined in the static trials during data collection or due to the characteristics of the sample populations. In situ studies have shown knee abduction and internal rotation increase ACL forces, particularly in the presence of smaller knee flexion angles.[18, 19] The lack of differences in internal rotation across sex and task does not lend a useful explanation of how internal rotation contributes to ACL loading during these dynamic tasks. It appears combining a lateral false step with a sidestep cut increases knee abduction, but not internal rotation, and may be a predictor of ACL injury during certain change-of-direction tasks.

Pollard et al.[26] were the first to investigate hip kinematics during sidestep cutting tasks and found females demonstrated greater hip adduction when compared to males. In addition, females have been found to display greater hip adduction during running compared to males.[9] Our results indicate hip adduction is significantly different during running and change-of-direction tasks and between females and males. It appears the amount of hip adduction associated with the task depends upon lateral placement of the foot such that the LFS and LFS-SSC maneuvers results in less hip adduction than running. In fact, the LFS-SSC maneuver results in less hip adduction than both the LFS and SSC tasks. Females demonstrated greater hip adduction compared to males, when collapsed across all four tasks. Frontal plane hip moments also differed across tasks with the LFS-SSC maneuver resulting in greater hip abduction moments compared to the RUN and LFS tasks. However, increased hip adduction angle for females was not concomitant with increased hip abduction moments compared to males. Females may rely on similar hip muscle activation patterns to males, but in a manner insufficient to control hip adduction. Differences in frontal plane hip position and loading between running and sidestep cutting tasks were expected; however, the lateral false step also appears to have a significant effect on frontal plane hip position and loading with respect to task and sex.

Our data did not support the presence of differences between females and males with respect to hip internal rotation; however, our results provide evidence that hip rotation differs across tasks. Both the SSC and LFS-SSC maneuvers resulted in greater internal hip rotation compared to the LFS and RUN tasks. While it may be that the amount of internal hip rotation is related to actually changing direction by 45º, the increases seen in the LFS-SSC compared to SSC may be the result of increasing the lateral placement of the foot. Hip external rotation moments were found to differ across tasks with the LFS-SSC resulting ingreater transverse plane hip loading when compared to RUN and LFS tasks. The lack of differences in hip external rotation moments between SSC and LFS-SSC tasks may be related to the similarities in hip internal rotation between these two tasks. Interestingly, females were found to demonstrate smaller hip external rotation moments compared to males. In may be the resulting differences in hip external rotation moments between females and males during the tasks in this study reflect variations in the contribution of the hip abductors to transverse and frontal plane loading of the hip as a function of sex. Females may rely on hip abductors to contribute more to the production of hip abduction moments than external rotation moments, thus explaining why there was an effect of sex on transverse plane hip loading but not frontal plane hip loading.

It has been suggested increasing internal knee adduction moments (external moments of knee abduction) may increase loading on the ACL during sidestep cutting. [21, 30] The presence of sex differences in knee frontal plane loading would in part provide explanation for why females are at greater risk of ACL injury compared to males. Although females have been found to have larger knee adduction moments compared to males for sidestep cutting tasks [21, 30] and running[9] our results did not support the presence of sex differences for knee adduction moments for running or any of the change-of-direction tasks. While it is not known how much knee adduction loading is necessary to disrupt the ACL in vitro, it may be that a specific population of athletes displays greater frontal plane loading at the knee when compared to others. Identifying the amount of frontal plane loading necessary to sufficiently injure the ACL in vivo and then determining the group of athletes with a propensity towards this amount of loading, may better help to identify those at risk of ACL injury.

Performing a lateral false step while running straight ahead significantly increased knee adduction moments during the initial part of stance. Our results also indicate the laterally directed step has a cumulative effect on frontal plane knee loading, as the LFS-SSC maneuver was associated with the greatest amount of knee adduction loading compared to all other tasks. Lloyd and Buchanan[16] have suggested non-contractile soft tissue structures are primarily responsible for providing support during abduction-adduction loading at the knee. Based upon our results, it appears the LFS-SSC maneuver increases loading on passive knee structures more than either sidestep cutting or a lateral false step maneuver. When considering that the ACL has been found to undergo greater loading under the application of external abduction torques[18, 19], we may conclude the ACL is at greater risk of injury during the LFS-SSC maneuver when compared to other change-of-direction tasks. Although we did not measure instances in which the ACL was actually injured, it is possible frontal plane knee loading may be more important in contributing to ACL injury than sagittal and transverse plane loading, as we did not detect differences in knee extension or external rotation moments across the tasks (Table 1).

While we did not find differences between females and males with respect to frontal plane hip loading, our results indicate males demonstrate greater hip external rotation moments than females in the initial part of stance during dynamic tasks. The differences in hip external rotation moments between females and males suggest males engage the hip extensors to a greater extent than females. The gluteus maximus muscle has been reported to function as a hip external rotator in addition to a hip extensor.[7] The demonstration of reduced hip external rotation moments for females across all tasks may indicate that either weakness of the gluteus maximus or an alternative recruitment pattern will result in a decreased ability to control the hip when running or changing direction compared to males. Perhaps a reduction in the control of transverse plane hip motion should be considered as a factor contributing to increased ACL injury risk.

The purpose of performing a sidestep cutting maneuver is to avoid an opponent and to transition between two distinct directions of travel when running. In order to produce an effective evasive movement, the athlete must perform the cutting maneuver powerfully to accelerate away from the opponent. Alternatively, increasing the deception associated with the sidestep cutting maneuver may also provide an advantage to the athlete, allowing for acceleration past their opponent. The addition of a lateral false step with a sidestep cut may be an effective means to facilitate deception. Notably, not only does the LFS contribute to greater knee and hip abduction compared to running, the effects on lower extremity loading are not significantly different than those for SSC. Our results further indicate that the addition of the laterally directed step in combination with the sidestep cut may be a significant factor with respect to frontal and transverse plane loading of the knee and hip respectively. In effect, increasing frontal plane motion at the onset of the sidestep cut may place the knee and hip in positions in which greater ACL loading may occur. We do not propose to discourage athletes from utilizing the lateral false step in combination with the sidestep cut, but suggest further investigations should determine why and to what extent this behavior is adopted by some athletes. Is it simply a matter of optimizing the outcomes of performing the maneuver, or that some athletes use the addition of frontal plane loading to increase how quickly and powerfully the maneuver can be performed if they cannot control sagittal plane deceleration effectively? In the presence of weak stabilizers through the transverse and frontal planes, this movement may increase the likelihood of knee injury.

One of the limitations to investigations of lower extremity biomechanics is related to the variability associated with transverse plane knee kinematics and kinetics. Although the approach taken to measuring these variables was similar to those of other researchers during three-dimensional analysis of movement, this may be one reason we were unable to detect differences in peak knee internal rotation and peak knee external rotation moments. Additionally, one explanation of dissimilarities in our findings with respect to frontal plane knee kinetics and kinematics when compared to other investigations may be the speed range used in the current study. Subjects in this study performed all trials at an average speed of 3.5 m/s, whereas the other researchers have adopted a speed parameter of 5.5-7.0 m/s. It may be differences between females and males in frontal plane knee position and loading during change-of-direction tasks only emerge at fast movement speeds. It also stands to reason we lacked sufficient power to detect small differences in knee biomechanics between males and females with only twenty-one subjects participating in this study.

In conclusion, variations of the sidestep cutting maneuver can produce equal or greater knee abduction angles and knee adduction moments than sidestep cutting in isolation. Frontal plane loading of the knee appears to be increased while frontal plane loading of the hip appears to be decreased when the athlete steps in a direction away from the intended change of direction in combination with the sidestep cut. The LFS appears to result in similar knee abduction and hip adduction positions in addition to similar knee adduction loading to those of sidestep cutting. Therefore, it is reasonable to further study this maneuver and consider its relationship to ACL injury. Females demonstrate significant differences in hip position and loading during running and change-of-direction tasks compared to males. The presence of increased hip adduction and reduced external rotation moments for females during change-of-direction tasks suggests differences in proximal control at the hip may help to explain why females are more susceptible to ACL injury than males. Further studies are necessary in order to determine what causative factors are associated with these differences in proximal control for females and what benefits might be associated with performing a change of direction with a laterally directed step in combination with a sidestep cut.

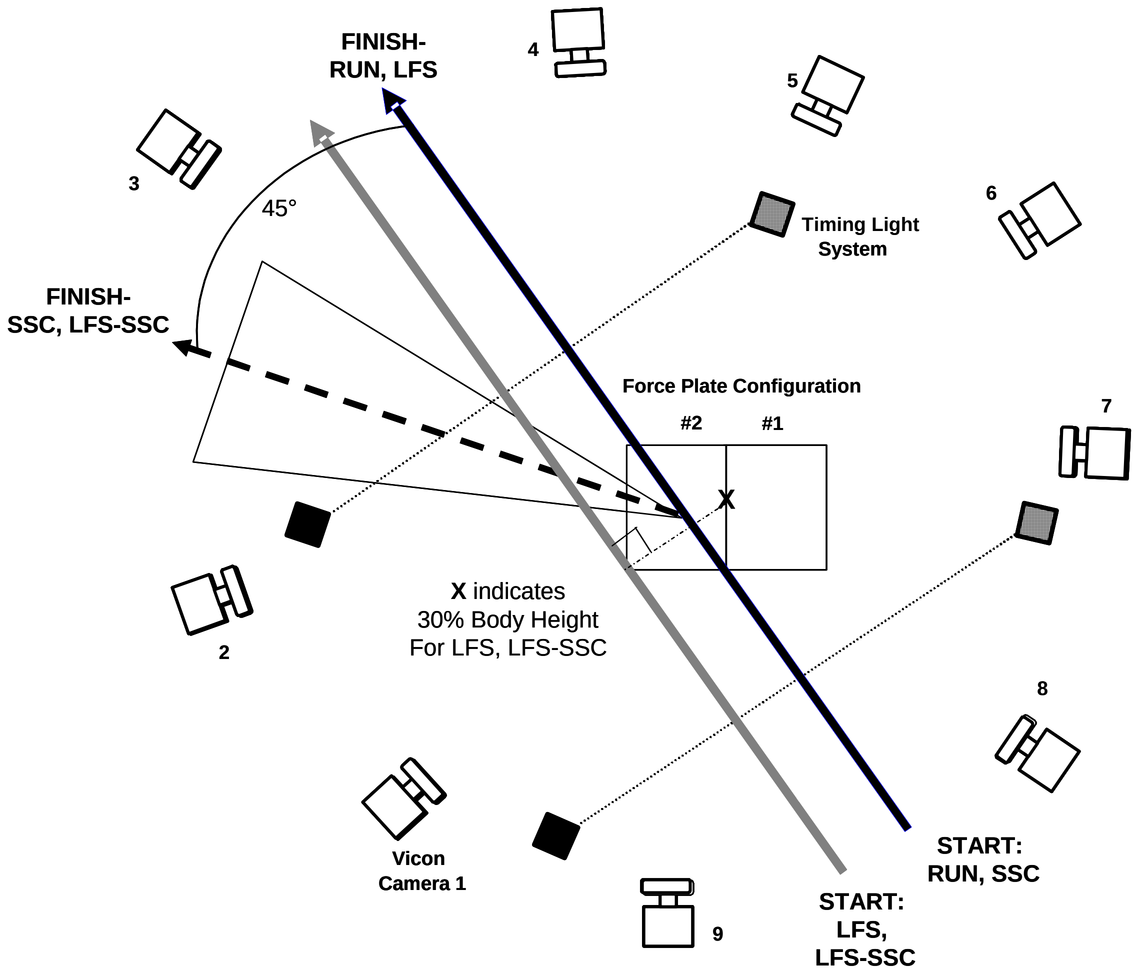


Figure 1. Laboratory set-up for data collection of running (RUN), lateral false step (LFS), sidestep cutting (SSC) and combination of lateral false step-sidestep cutting (LFS-SSC) maneuvers. Distance from the running lines for LFS and LFS-SSC maneuvers was determined to be 30% of body height and measured at a distance perpendicular to the direction of travel for both conditions. Change-of-direction angle for SSC and LFS-SSC conditions was set to 45 ± 10º. Tape was applied to the floor to indicate the different paths of travel for all conditions in addition to the acceptable range of 35-55º for the SSC and LFS-SSC conditions.

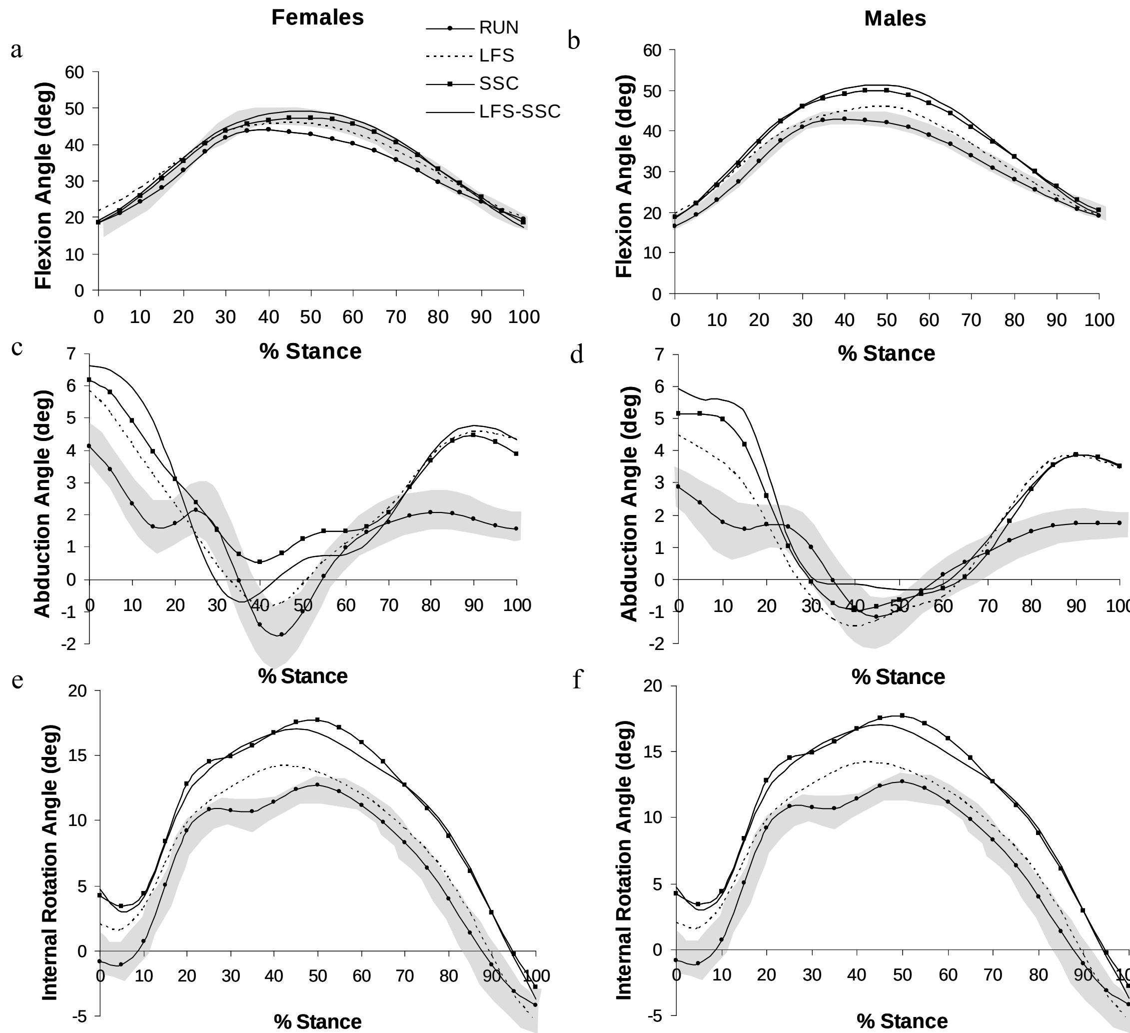


Figure 2. Ensemble-average joint angles for knee rotations: flexion-extension (a = females, b = males), abduction-adduction (c = females, d = males), internal-external rotation (e = females, f = males) during stance phase for running (RUN), a lateral false step (LFS), sidestep cutting (SSC) and a combination of lateral false step-sidestep cutting (LFS-SSC). Shaded areas represent ± 1SD for RUN. Positive values indicate flexion, abduction, internal rotation.

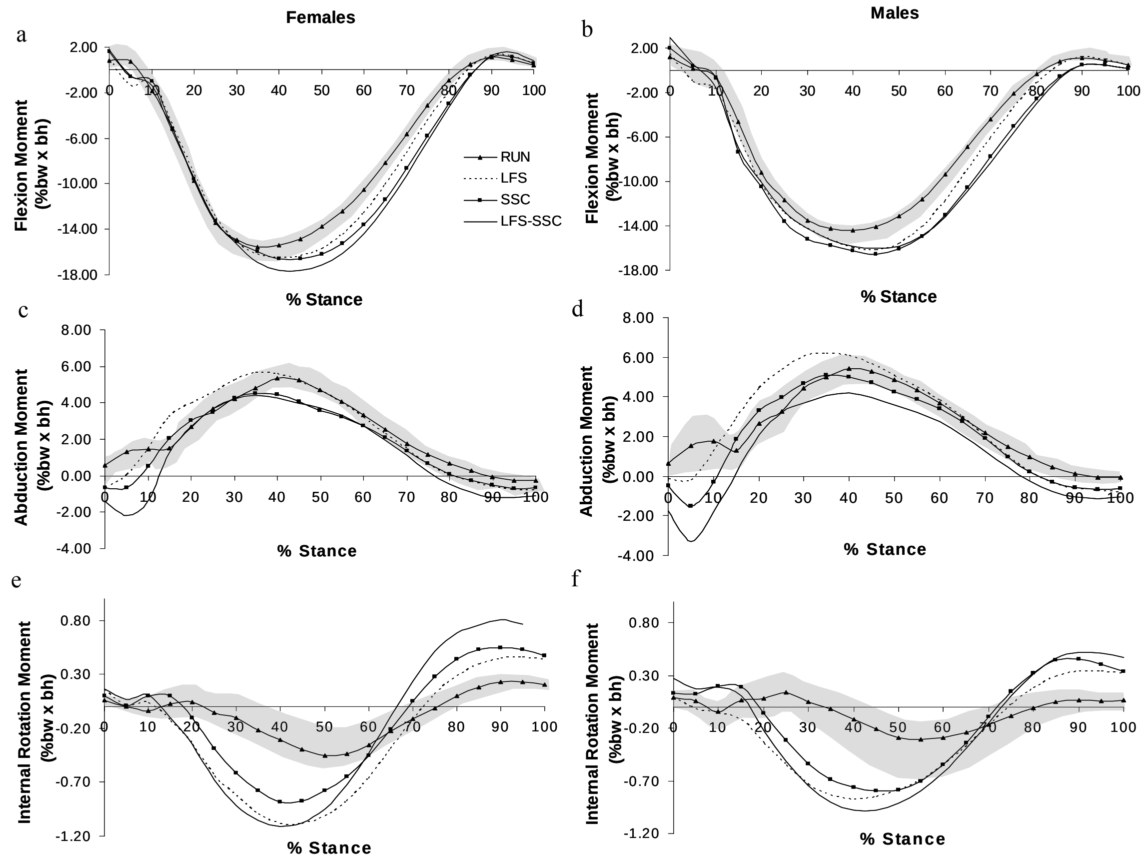


Figure 3. Ensemble-average internal joint moments for knee rotations: flexion-extension (a = females, b = males), abduction-adduction (c = females, d = males), internal-external rotation (e = females, f = males) during stance phase for RUN, LFS, SSC and LFS-SSC. Shaded areas represent ± 1SD for RUN. Positive values indicate flexion, abduction, internal rotation. Abbreviations are the same as in Figure 2.

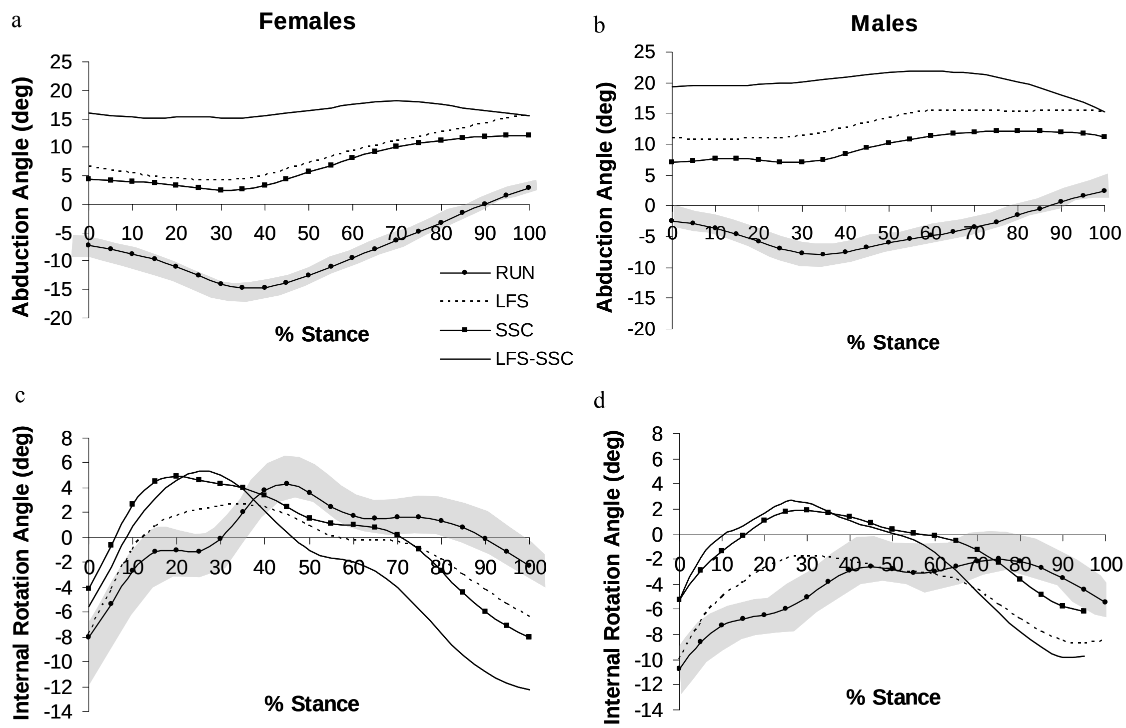


Figure 4. Ensemble-average joint angles for hip rotations: abduction-adduction (a = females, b = males), internal-external rotation (c = females, d = males) during stance phase for RUN, LFS, SSC and LFS-SSC. Shaded areas represent ± 1SD for RUN. Positive values indicate flexion, abduction, internal rotation. Abbreviations are the same as in Figure 2.

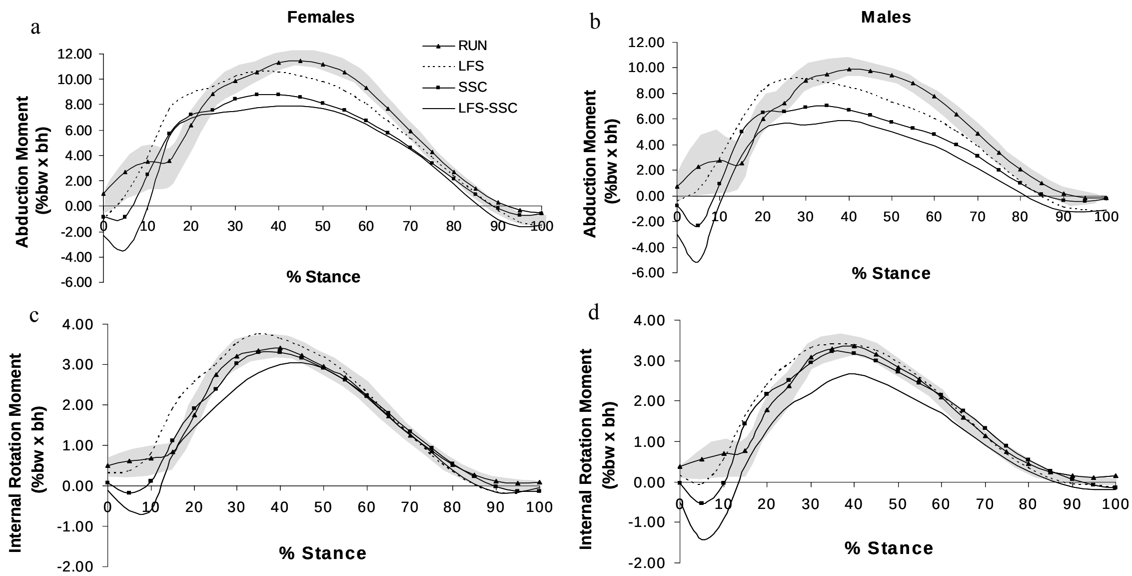


Figure 5. Ensemble-average internal joint moments for hip rotations: abduction-adduction (a = females, b = males), internal-external rotation (c = females, d = males) during stance phase for RUN, LFS, SSC and LFS-SSC. Shaded areas represent ± 1SD for RUN. Positive values indicate flexion, abduction, internal rotation. Abbreviations are the same as in Figure 2.

TABLE 1

Condition-based comparisons of knee and hip angles and moments[a]

Variable	Condition: RUN	LFS	SSC	LFS-SSC	P-value
Peak Angle					
Knee abduction	3.62 ± 0.52	5.57 ± 0.54[c]	6.12 ± 0.50[c]	7.07 ± 0.53[cde]	< 0.001
Knee internal rotation	7.44 ± 1.25	8.36 ± 1.39	8.79 ± 1.55	9.44 ± 1.50	0.26
Hip adduction[b]	7.26 ± 0.51	-7.86 ± 0.76[c]	- 5.67 ± 0.87[c]	-16.62 ± 0.80[cde]	< 0.001
Hip internal rotation[b]	- 4.15 ± 1.06	- 3.12 ± 1.64	0.41 ± 1.55[c]	1.03 ± 1.92[cd]	< 0.001
Peak Moment					
Knee extension	5.40 ± 0.53	4.71 ± 0.57	5.64 ± 0.94	5.02 ± 0.68	0.48
Knee adduction	0.13 ± 0.12	1.16 ± 0.18 [c]	2.04 ± 0.22[c]	3.93 ± 0.06[cde]	< 0.001
Knee external rotation	0.13 ± 0.03	0.21 ± 0.03	0.10 ± 0.04	0.17 ± 0.07	0.25
Hip abduction	5.50 ± 0.50	5.81 ± 0.80	4.30 ± 0.83	2.74 ± 0.76[cd]	< 0.001
Hip external rotation[b]	- 0.16 ± 0.06	0.05 ± 0.11	0.56 ± 0.11	1.31 ± 0.17[cd]	< 0.001

[a] Values represent mean ± SD (N = 21) of knee and hip peak joint angles (º) and internal moments (%bw x bh) from initial contact to 30º of knee flexion during stance phase, collapsed across sex, for each of the four running and change-of-direction tasks (running = RUN, lateral false step = LFS, sidestep cutting = SSC, and combination of lateral false step and sidestep cutting = LFS-SSC). bw = body weight; bh = body height

[b] negative values indicate that the joint angle or moment did not cross zero in the corresponding direction.

[c] significantly different than RUN (Tukey's HSD $p < 0.05$)

[d] significantly different than LFS (Tukey's HSD $p < 0.05$)

[e] significantly different than SSC (Tukey's HSD $p < 0.05$)

TABLE 2
Sex-based differences in knee and hip angles and moments[a]

Variable	Sex: Females	Sex: Males	P-value
Peak Angle			
Knee abduction	6.09 ± 0.67	5.10 ± 0.70	0.32
Knee internal rotation	9.95 ± 1.78	7.06 ± 1.87	0.19
Hip adduction	-3.74 ± 0.83	-7.70 ± 0.82	0.04
Hip internal rotation	0.53 ± 1.88	2.39 ± 1.97	0.50
Peak Moment			
Knee extension	4.92 ± 0.80	5.47 ±0.84	0.64
Knee adduction	1.60 ± 0.24	1.90 ± 0.26	0.42
Knee external rotation	0.19 ± 0.04	0.12 ± 0.04	0.22
Hip abduction	4.86 ± 0.86	4.31 ± 0.90	0.67
Hip external rotation	0.22 ± 0.12	0.66 ± 0.13	0.02

[a] Values represent mean ± SD (Females = 11, Males = 10) of knee and hip peak joint angles (º) and internal moments (%bw x bh), collapsed across condition, from initial contact to 30º of knee flexion during stance phase for females and males. bw = body weight; bh = body height
[b] negative values indicate that the joint angle or moment did not cross zero in the corresponding direction.

REFERENCES

1. Agel J, Arendt EA, Bershadsky B. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med.* 2005;33(4):524-530.PMID: 15722283 Provider: OCLC

2. Andrews JR, McLeod WD, Ward T, Howard K. The cutting mechanism. *Am J Sports Med.* 1977;5(3):111-121.PMID: 860773 Provider: OCLC

3. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. *Am J Sports Med.* Nov-Dec 1995;23(6):694-701.8600737

4. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* Jun 2000;23(6):573-578.PMID: 10875418; Medline: 20331673 Provider: OCLC

5. Cross MJ, Gibbs NJ, Bryant GJ. An analysis of the sidestep cutting manoeuvre. *The American journal of sports medicine.* May-Jun 1989;17(3):363-366.PMID: 2729486; Medline: 89270834 Provider: OCLC

6. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech.* 1996;29(9):1223-1230.PMID: 8872282 Provider: OCLC

7. Delp SL, Hess WE, Hungerford DS, Jones LC. Variation of rotation moment arms with hip flexion. *J Biomech.* 1999;32(5):493-501.PMID: 10327003 Provider: OCLC

8. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med.* Mar 2004;32(2):477-483.PMID: 14977677 Provider: OCLC

9. Ferber R, Davis IM, Williams DS. Gender differences in lower extremity mechanics during running. *Clin Biomech.* 2003;18:350-357.PMID: 12689785 Provider: OCLC

10. Ford KR, Myer GD, Toms HE, Hewett TE. Gender differences in the kinematics of unanticipated cutting in young athletes. *Med Sci Sports Exerc.* 2005;37:124-129.PMID: 15632678 Provider: OCLC

11. Golden GM. *The effect of altering the width of a single step of a running stride on knee joint kinematics and kinetics* [In Lieu of Thesis, unpublished manuscript]. Corvallis: Nutrition and Exercise Sciences, Oregon State University; 2005.

12. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, Garrick JG, Hewett TE, Huston L, Ireland ML, Johnson RJ, Kibler WB, Lephart S, Lewis JL, Lindenfeld TN, Mandelbaum BR, Marchak P, Teitz CC, Wojtys EM. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *The Journal of the American Academy of Orthopaedic Surgeons.* May-Jun 2000;8(3):141-150.PMID: 10874221; Medline: 20334910 Provider: OCLC

13. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng.* 1983;105(2):136-144.PMID: 6865355 Provider: OCLC

14. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, van den Bogert AJ, Paterno MV, Succop P. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* 2005 2005;33(4):492-501.PMID: 15722287 Provider: OCLC

15. Kirkendall DT, Garrett WE. The anterior cruciate ligament enigma. Injury mechanisms and prevention. *Clinical orthopaedics and related research.* 2000;372:64-68.PMID: 10738415 Provider: OCLC

16. Lloyd DG, Buchanan TS. A model of load sharing between muscles and soft tissues at the human knee during static tasks. *J Biomech Eng.* Aug 1996;118(3):367-376.PMID: 8872259 Provider: OCLC

17. Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of knee joint motion patterns between men and women in selected athletic tasks. *Clinical biomechanics.* Jun 2001;16(5):438-445.PMID: 11390052; Medline: 21285965 Provider: OCLC

18. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, J.l. S. Combined knee loading states that generate high anterior cruciate ligament forces. *Orhop Res. Society.* 1995;13:930-935.PMID: 8544031 Provider: OCLC

19. Markolf KL, Gorek JF, Kabo JM, Shapiro MS. Direct measurement of resultant forces in the anterior cruciate ligament. An in vitro study performed with a new experimental technique. *J Bone Joint Surg Am).* 1990;72:557-567.PMID: 2324143 Provider: OCLC

20. McLean SG, Huang X, Su A, Van Den Bogert AJ. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clin Biomech.* 2004;19(8): 828-838.PMID: 15342155 Provider: OCLC

21. McLean SG, Huang X, van den Bogert AJ. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech.* 2005;20(8):863-870.PMID: 16005555 Provider: OCLC

22. McLean SG, Lipfert SW, van den Bogert AJ. Effect of gender and defensive opponent on the biomechanics of sidestep cutting. 2004.PMID: 15179171 Provider: OCLC

23. McLean SG, Myers PT, Neal RJ, Walters MR. A quantitative analysis of knee joint kinematics during the sidestep cutting maneuver. Implications for non-contact anterior cruciate ligament injury. *Bull Hosp Jt Dis.* 1998;57(1):30-38.PMID: 9553700 Provider: OCLC

24. McLean SG, Neal RJ, Myers PT, Walters MR. Knee joint kinematics during the sidestep cutting maneuver: potential for injury in women. *Med Sci Sports Exerc.* 1999;31(7):959-968.PMID: 10416556 Provider: OCLC

25. McNair PJ, Marshall RN, Matheson JA. Important features associated with acute anterior cruciate ligament injury. *The New Zealand medical journal.* Nov 14 1990;103(901):537-539.PMID: 2243642; Medline: 91056954 Provider: OCLC

26. Pollard CD, Davis IM, Hamill J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech.* 2004;19(10):1022-1031.PMID: 15531052 Provider: OCLC

27. Powers CM. The Influence of Altered Lower-Extremity Kinematics on Patellofemoral Joint Dysfunction: A Theoretical Perspective. *J Orthop Sports Phys Ther.* 2003;33:693-646

28. Romani W, Patrie J, Curl LA, Flaws JA. The correlations between estradiol, estrone, estriol, progesterone, and sex hormone-binding globulin and anterior cruciate ligament stiffness in healthy, active females. *J Womens Health.* 2003;12(3):287-298.PMID: 12804359 Provider: OCLC

29. Shultz SJ, Kirk SE, Johnson ML, Sander TC, Perrin DH. Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Med Sci Sports Exerc.* 2004;36(7):1165-1174.PMID: 15235320 Provider: OCLC

30. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. *Clin Biomech.* 2006;21:41-48.PMID: 16209900 Provider: OCLC

31. Vaughan CL, Davis, B. L., O'Conner, J. C. *Dynamics of Human Gait*. 2nd ed. Cape Town, South Africa: Kiboho Publishers; 1999.

32. Wascher DC, Markolf KL, Shapiro MS, Finerman GA. Direct in vitro measurement of forces in the cruciate ligaments. Part I: The effect of multiplane loading in the intact knee. *J Bone Joint Surg Am.* 1993;75:377-386.PMID: 8444916 Provider: OCLC

33. Winter DA. *Biomechanics and motor control of human movement*. Second ed. Toronto: Wiley-Interscience; 1990.

Relationship of static lower extremity alignment and hip strength to frontal plane knee position and loading during running and rapid change-of-direction tasks

Grace M. Golden, Michael J. Pavol, Mark A. Hoffman

INTRODUCTION

The prevalence and consequences of non-contact anterior cruciate ligament (ACL) injuries in the athletic population motivate researchers to determine which risk factors predispose individuals to ACL disruption. Each year, an estimated 1 in 3000 individuals sustains an ACL injury in the United States[21], with approximately 50,000 ACL reconstructions performed annually.[11] At the collegiate level of athletic competition, 32% of knee injuries in the sports of basketball and soccer involve ACL injury.[3] Additionally, the gender bias, in which females are more likely to suffer this injury than males[1, 3], continues to elude comprehensive explanation. Regardless of sex or sport, the consequences of ACL injury are significant, including costs associated with surgical reconstruction and management, time loss from participation, and long term sequelae, as Maletius and Messner[28] report that over 80% of individuals who suffer ACL injury will develop knee osteoarthritis.

High risk maneuvers in which the ACL is susceptible to injury include rapid changes of direction[2, 4], abrupt stopping[15, 31], and single- or double-leg jump landings[15, 31]. The deleterious knee position associated with these maneuvers is characterized by internal rotation and abduction when the knee is relatively extended.[25] Intrinsic injury risk factors may be related to ACL geometry and structural properties, femoral notch width, body mass index, lower extremity alignment (i.e. Q-angle), and hormonal differences.[16, 21, 38] Inasmuch as some intrinsic risk factors can be identified but not prophylactically modified (e.g. ACL geometry, femoral notch width), others such as lower extremity alignment offer potential for screening and intervention. The notion of a relationship between Q-angle and frontal plane knee position during dynamic tasks is inviting, as clinicians can efficiently perform measurement of static limb alignment.

Differences in lower extremity static posture between males and females have been reported, however, its relationship to dynamic knee and hip positions is still unclear. Quadriceps angle (Q-angle) represents one appraisal of lower extremity alignment and is measured as the angle created by two bisecting lines: 1.) from midpoint of the patella to the anterior superior iliac spine and 2.) from the midpoint of the tibial tuberosity to the midpoint of the patella. Females have been found to have a greater Q-angle compared to males.[20] Individuals with symptomatic knee pain have also been found to have bilateral asymmetries in Q-angle which may thus be a predictor of patellofemoral knee pain.[32] Skeletal contributions to differences in Q-angle are not clearly understood, however. Powers[36] suggests increased knee abduction augments Q-angle as a result of a combination of increased femoral adduction, tibial abduction, and femoral internal rotation. This combination of segmental alignment may affect patella displacement relative to the anterior superior iliac spine and tibial tuberosity. Quadriceps strength has a direct effect on patella position, supported by the research of Byl et. al[5], who found an inverse relationship between knee extensor torque and Q-angle. However, to date, there has not been exploration of the relationship between hip strength and Q-angle. Ireland[22] asserts that muscular action at the hip affects the knee, particularly for the female athlete. In fact, improving hip abductor strength has been shown to have a positive effect in reducing iliotibial band syndrome in runners[12] This may be due to the primary and secondary roles of the gluteus medius at the hip. Mansour and Pereira[29] determined that the anterior and middle fibers of the gluteus medius function as hip abductors during portions of the gait cycle and additionally, the anterior fibers produce an internal rotation moment when the hip is flexed. The middle and posterior portions of the gluteus medius were found to produce an external hip rotation moment when the hip was relatively extended. Similarly, Delp et. al[10] determined that four different portions of the gluteus medius have internal rotation moment arms when hip flexion position increases from 0-90º. This supports the role of the gluteus medius not only as a hip abductor, but as a controller of hip rotation and therefore potentially important in affecting frontal plane knee position. Specifically, strengthening of the gluteus medius muscle may improve frontal and transverse plane position of the femur which will directly influence Q-angle.

Despite intriguing support of the influence of hip strength on lower extremity alignment, the relationship between Q-angle and dynamic alignment of the limb is tenuous. While Pantano et al.[34] did not find association between Q-angle and dynamic alignment of the limb during a single-leg squat, others have suggested that static Q-angle contributes to the predictability of knee abduction position during single-leg landings.[16] Recently, in an investigation of eccentric hip abductor strength and knee joint kinematics during jump landing tasks, Jacobs and Mattacola[23] found a significant inverse relationship between female peak eccentric hip strength and peak knee abduction angle, even in the absence of differences between male and female peak eccentric hip abduction strength (normalized to body weight). Specifically, females demonstrating larger eccentric peak torques were found to have lower peak knee abduction angles during landings. In a similar investigation, researchers found a mild relationship between isometric hip abduction strength and knee abduction during jump landings for females, but not males.[24] What is not known is whether hip strength has a direct relationship with static lower extremity alignment and dynamic limb alignments in the frontal plane during change-of-direction tasks. More study is needed to solidify this potential relationship during rapid change-of-direction tasks in addition to determining whether hip strength is related to frontal plane knee position.

Therefore, the purpose of this study was several-fold. The first purpose was to compare standing Q-angle, hip abduction strength, and the position of peak torque production during eccentric hip abduction in collegiate female and male athletes. The second was to determine whether either standing Q-angle or hip abduction strength is related to frontal plane peak knee position and loading during running and rapid change-of-direction tasks. The subjects in this study had previously been found to have no sex-related differences in peak knee abduction or knee adduction moments during the same running and change-of-direction tasks represented in this current investigation.[14] The rapid change-of-direction tasks of interest in this study included a lateral false step maneuver (LFS), sidestep cutting (SSC), and a combination of a lateral false step maneuver and sidestep cutting (LFS-SSC). The LFS maneuver represented a step within a running stride directed laterally at a specific percentage of body height (30%), followed by an immediate return to the original running path.[13] Sidestep cutting involved redirecting the direction of travel abruptly at an angle of 45º.[30, 35, 40] The combination of the two maneuvers resulted when the step preceding the sidestep cut was exaggerated away from the change of direction by 30% of body height. The first hypothesis was that female basketball and soccer athletes would exhibit greater standing Q-angles and lesser hip abduction strength (normalized to body weight) compared to males from matched sports. Additionally, it was expected there would be no differences between males and females in the angle at which the peak torque was produced. Secondly, it was hypothesized that athletes demonstrating greater standing Q-angle measurements would also exhibit greater peak knee abduction during running and rapid change-of-direction tasks. Finally, it was theorized that athletes having greater eccentric hip abduction strength would have reduced internal peak knee adduction moments during running and rapid change-of-direction tasks. All comparisons of peak knee kinematics and kinetics were constrained to the initial portion of stance in which the knee was in less than 30º of flexion.

METHODS

Twenty-one subjects (11 females: 19.8 ± 1.5 yrs, 175.8 ± 6.3 cm, 69.9 ± 6.6 kg; 10 males: 20.7 ± 2.5 yrs, 188.6 ± 15.9 cm, 89.8 ± 21.7 kg) , basketball and soccer athletes from four college and university programs, were enrolled in the study. During initial screening, subjects completed demographic and health history questionnaires. Exclusion criteria included a past history of any reconstructive surgical procedure for the treatment of a lower extremity injury within the past twelve month period. All subjects were required to have an intact ACL. Subjects also could not be currently receiving supervised medical care for an injury to the upper or lower extremity preventing their participation in regular conditioning or practice sessions nor could they have suffered an injury to the lower extremity resulting in time loss from participation exceeding three weeks within the past six months. Additionally, female subjects were screened for a regular menstrual cycle (28-32 days[37, 39]) for the past 3 months and participated in the testing session within 1-9 days of the start of menses, i.e. the follicular phase (6.5 ± 1.8 days). All participants were of equivalent sport experience (females: 8.45 ± 1.97 yrs, males: 7.00 ± 1.76 yrs; $p > 0.05$) and participating in either off-season conditioning (basketball) or a non-traditional competitive season (soccer). Recruitment and participation of subjects was approved by a University Institutional Review Board for the Protection of Human Subjects and all subjects provided informed consent prior to participation.

Standing Q-angle was measured with a plastic goniometer on the right limb. Subjects were barefoot and positioned in the Rhomberg position (medial malloeli touching, standing with relaxed quadriceps).[5] The center of the patella, middle of the tibial tuberosity and anterior superior iliac spine (ASIS) were identified and marked with a felt pen. The fulcrum of the goniometer was placed at the patella midpoint and the moving and stationary arms of the goniometer were aligned with the ASIS and tibial tuberosity respectively. Three measurements were taken and the average used for analysis. Intertester reliability for measurement of standing Q-angle was determined from a preliminary study (N = 6) and found to be excellent ($ICC_{2,1}$ $r = 0.93$)

Eccentric hip abduction strength data were collected for the right limb with a Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Shirley, NY), interfaced with a Biopac MP100 Data Collection system (Biopac Systems, Inc., Goleta, CA). Torque and corresponding hip angle data from the dynamometer were stored and processed on a desktop computer with Acqknowledge Waveform Analysis

Software (Version 3.7). Eccentric hip abduction strength was measured in a standing position with the axis of rotation of the dynamometer positioned at the ASIS. Threshold torque was established at 272Nm for all subjects. Subjects performed 2 sets of 5 repetitions. The first set was designated as a practice set and was followed by a 2 minute rest period before completion of the test set. Eccentric abduction strength was measured from a starting position of 30º of hip abduction to a final position of 0º of hip abduction at a velocity of 120º/sec. The measured torque for each test trial was smoothed and the peak torque was extracted. Peak torque was then normalized to subject's body mass and body height (Nm/kg/m)
The angle of peak torque development was determined from the measured hip torque and angle data in the test set for eccentric hip strength, utilizing "the find peak" function in the Acqknowledge Waveform Analysis Software.

Kinematic and kinetic data acquisition during running and change-of-direction tasks were performed with a Vicon 612 optical motion capture system utilizing 9 high-resolution MCam2 digital cameras (Vicon, Lake Forest, CA), interfaced with two Bertec Model 4060-08 force plates (Bertec, Columbus, OH) that were configured to function as a single force plate. Sixteen markers (9mm diameter) were affixed bilaterally on the lower extremity with double sided tape in order to identify body segment positions during static and dynamic trails. A customized marker set, based upon the work of Vaughan et al.[41] was utilized for this study (5th metatarsal, heel, ankle, tibia, knee, thigh, anterior superior iliac spine, and posterior superior iliac spine). Kinematic data were collected at 120Hz and kinetic data were collected throughout foot contact at 1080 Hz.

Figure 1 illustrates the laboratory set up for the running and change-of-direction tasks. In order to identify the path of travel for RUN and SSC conditions, yellow tape was affixed to the floor in a line crossing the center of the force plate. A second, parallel, blue tape line was placed on the floor, crossing the left corner of the force plate to indicate the path of travel for the LFS and LFS-SSC conditions. A divergent tape line, originating from the force plates, was placed at an angle of 45º to the left of the other lines to identify the pathway for subjects to successfully perform the SSC and LFS-SSC change-of-direction maneuvers. Colored adhesive tape was applied on the force plate to indicate a distance of 30% of body height for LFS and LFS-SSC conditions. This distance was measured perpendicularly to the right from the center point of the tape line crossing the left corner of the force plate. Cameras were positioned so each marker was visible by multiple cameras when the right foot was in contact with the force plate. Each trial consisted of a running a distance of approximately 10m from start to finish.

On the day of testing, subjects reported to the testing laboratory. Height (cm) was measured (with shoes on) with a wall mounted stadiometer. Weight (lbs) was recorded with a floor scale and later converted to kg. After measurement of SQA, subjects performed a 5 minute warm-up by running on a treadmill (2.5-3.1 $m \cdot s^{-1}$). During this time, adhesive tape was applied to the force platform at a location indicating a distance equivalent to 30% of body height from the line of travel for the LFS and LFS-SSC conditions.

After the warm-up period, measurements of standing Q-angle were obtained. Next, subjects were allowed 3-5 practice trials for each of the running and change-of-direction tasks in the order or RUN, LFS, SSC, LFS-SSC. These practice trials enabled researchers to determine the most effective starting location so that the running stride prior to force plate contact was fluid and consistent, and improved subject familiarization with the specific tasks. Throughout the practice and testing trials, the speed of the subject was monitored through the use of timing lights placed 1.5 m to the rear and front of the force plates, at a total distance of 3 m apart. Trials were considered acceptable when speed equated to 3.5 $m \cdot s^{-1} \pm 10\%$ (3.15 - 3.85 $m \cdot s^{-1}$). Recovery was accommodated by having subjects walk back to the starting location between each trial. Verbal instructions for running speed and technique were standardized for all subjects. Subjects were directed to perform all trials without a reduction of speed into the RUN, LFS, SSC, and LFS-SSC maneuvers, discouraging leaping or hesitating prior to force plate contact. Testing trials were discarded if the subject made foot contact on the force plates with the left foot, failed to make contact with the force plates with the entire right foot, missed the LFS and LFS-SSC tape mark on the force plate, initiated the LFS or LFS-SSC maneuver by crossing the left foot to the right of the directional guide placed on the floor, or completed the SSC or LFS-SSC cut outside a range of 45º ± 10º. Colored tape was placed on the floor to indicate the boundaries of 35-55º.

Following familiarization of running and change-of-direction tasks, markers were affixed to the subjects. Each subject wore form-fitting spandex shorts of a dark color and a pair of their own team-issued running shoes for data acquisition trials. Foot markers were placed on the outside of the running shoe. A static trial was collected for calibration of segmental orientation, with the subject positioned in anatomical neutral (arms at 90º abduction) while standing on the force plates. Ten successful trials were recorded for each of the four conditions in the order of: RUN, LFS, SSC, and LFS-SSC. During each trial, kinematic and kinetic data were measured for the subject's right limb throughout contact with the force plate during stance phase. Subsequent to the completion of all running and change-of-direction trials, foot length (shoe length) and pelvis, knee, and ankle width were measured with a joint anthropometer (Lafayette Instrument Company, Lafayette, IN). Three measurements were obtained for each measurement by the same researcher and averaged for data analysis. Testing of eccentric hip abduction strength was conducted after acquisition of three-dimensional kinematics and kinetics. All data were collected in a single testing session.

Three degrees of freedom were assigned to the major joints of interest with the corresponding rotations of flexion-extension, abduction-adduction, and internal-external rotation for the hip, knee, and ankle joints. Raw data for all dynamic trials were used to determine the three-dimensional paths of the markers and cubic spline interpolation was performed to fill small gaps (less then 50ms) with Vicon Workstation software. Kinematic data were filtered using a fourth-order zero-lag Butterworth low-pass filter with a cut-off frequency of 16Hz, as determined through a residual analysis.[43] Marker positions from the static, standing trial and anthropometric measures were used to establish a set of transformations for determining joint center locations and body segment orientations. These transformations were used to determine the paths of the joint centers and the orientations of the body segments over the entire stance phase of running. From these, the joint angles (degrees) at the stance knee were determined by the convention of Grood and Suntay[17]. Knee internal rotation angles during the static trial were defined to be zero. The computed kinematics were also combined with the force plate data to determine the internal joint moments acting at the knee during stance using 3-dimensional inverse dynamics methods. Segmental inertial parameters were determined as outlined by de Leva for female and male data.[9] Joint moments were expressed with respect to the joint rotation axes. Joint moments (Nmm) were normalized to body height (mm) and body weight (N) and reported as a percentage of body height (bh) and body weight (bw). A customized Matlab program (Student version 7, Mathworks, Natick, MA) was created to extract peak joint angles and moments from initial contact to 30º of knee flexion during stance for all dynamic trials. An average of the peak values from the ten trials of each condition by each subject was used for statistical analysis.

Independent t-tests were used to analyze differences between females and males for standing Q-angle (SQA; º), peak eccentric hip abduction strength (ECC; Nm/(kg·m)), and the angle of peak torque production (POS, º). Separate linear regression analyses were used to determine the following relationships for each of the four running and change-of-direction tasks 1.) peak knee abduction angle as a function of SQA, sex, and their interaction, 2.) peak knee adduction moment as a function of SQA, sex, and their interaction , 3.) peak knee abduction angle as a function of ECC, sex, and their interaction, and 4.) peak knee adduction moment as a function of ECC, sex, and their interaction. For all statistical analyses, an a priori alpha level of 0.05 was used to test for statistical significance. Analysis was performed with SPSS software (version 13.0, Chicago, IL).

RESULTS

There was no evidence of significant differences in SQA ($p = 0.16$), ECC ($p = 0.44$), or POS for the production of peak torque ($p = 0.78$) between female and male basketball and soccer players (Table 1).

Peak knee abduction angle in the initial part of stance was not found to be associated with SQA. sex, or the interaction between sex and standing Q-angle for the conditions of RUN, SSC, or LFS-SSC (Table 2). Although the regression model for peak knee abduction angle as a function of SQA, sex, and their interaction was found to be significant ($p = 0.03$), none of the explanatory variables were found to be statistically significant (SQA, $p = 0.33$; sex, $p = 0.23$; sex x SQA, $p = 0.90$). Peak knee adduction moment in the initial part of stance was not found to have a significant relationship with SQA, sex, or the interaction between sex and SQA for any of the four conditions (Table 2).

Peak knee abduction angle in the initial part of stance was found to have a significant relationship with peak eccentric hip abduction strength, sex, and their interaction for the LFS condition ($p = 0.01$). Individually, neither hip strength nor sex contributed significantly to this relationship, however the interaction between sex and hip strength approached significance ($p = 0.05$). Peak knee abduction angles and peak knee adduction moments in the initial part of stance were not found to be associated with peak eccentric hip abduction strength, sex, or their interaction for the conditions of RUN, SSC, or LFS-SSC (Table 3).

DISCUSSION

Our results for standing Q-angle did not confirm the existence of differences between males and females. Although females in this study demonstrated a standing Q-angle of 2º more than males, the difference was not significantly different. The standing Q-angle across males and females in this study were similar to values reported by Byl et al.[5] but on average were 5.45º lower than data of other investigators.[19, 20, 34, 44] Other researchers have reported sex differences of 3.3-4.6º [19, 20, 44]. It may be that the athletes used in this study represented a more homogenous population than those in other studies and may in part, explain the lack of agreement with regards to an effect of sex on standing Q-angle.

Data from this investigation suggest females and males do not differ in eccentric hip abduction strength when normalized to body mass and body height. These results support the works of Jacobs and Mattacola[23] and Claiborne et al.[8] who normalized peak hip abduction torque to body weight. However Cahalan et al.[6] found that overall, males demonstrate greater normalized hip abduction strength compared to females when evaluated in a standing position. It is not clear whether these measurements appraised concentric or eccentric muscle actions. Hollman et al.[19] and Leetun et al.[27] reported isometric hip abduction strength between males and females to differ by 5.6% and 3.4% respectively when normalized to body weight. Additionally, Jacobs et al.[24] found males to have greater isometric hip abduction strength (normalized to body height and body weight) compared to females. The disagreement across studies may indicate that differences in concentric and isometric abduction hip strength may not be correlated to similar differences in eccentric measurements. Therefore the differences between males and females may be related to the mode of evaluating hip strength. Additionally, it is not clear if hip position, seated vs. standing, contributes to the lack of agreement across studies, as there were differences in the position the torque measurement was made. Model-generated moment arms of hip muscles have been found to be affected by relative hip flexion.[10] In particular, the gluteus medius functions as a hip rotator and therefore it is possible that its relative contribution to hip abduction strength may change depending upon the amount of hip flexion associated with the torque measurement. Otherwise, it may be that our sample population was relatively homogenous with respect to hip strength measurements since they were all collegiate athletes with similar years of sport maturation.

While researchers have proposed that lower limb structural alignment will affect knee abduction[16, 22], the results from this study do not provide evidence of a relationship between standing Q-angle and peak knee abduction in early stance for either running or dynamic change of direction tasks. Pantano et al.[34] were also unable to detect a relationship between standing Q-angle and dynamic knee abduction during a single-leg squat. Griffin et al.[15] and Moul[33] suggest standing Q-angle may be a possible risk factor for ACL injury. Although clinical assessment of this measurement is relatively easy, we do not suspect it is a good predictor of lower extremity alignment during dynamic tasks or of potential ACL injury risk. It has been shown that standing Q-angle is modifiable due to isometric quadriceps contraction[5, 18, 26] It stands to reason that the relationships between hip and knee positions and muscle loading during dynamic tasks are inter-related. These relationships depend upon multiple factors, such that structural alignment in the absence of active limb loading lacks sensitivity to predict lower extremity position in the frontal plane during dynamic tasks.

Recent evidence demonstrates hip abduction strength to be related to frontal plane knee position in single- and double-leg jump landings.[23, 24, 42] Contrary to these reports, Hollman et al.[19] found that the hip abduction/adduction isometric force ratio was positively correlated with longitudinal arch angle, but not with standing Q-angle. Although our data did not represent a hip force ratio, the results suggest eccentric hip abduction strength was not related to frontal plane knee motion during dynamic tasks. Powers[36] theorizes proximal factors, such as hip strength, may help to explain differences in lower extremity alignment, such as knee abduction, during weight bearing activities. It may be that the lack of a relationship between knee abduction and hip strength in this study is confounded by the fact that these same subjects did not demonstrate differences in frontal plane knee angle as an effect of sex.[14] Conversely, these females were found to have greater hip adduction for the conditions of LFS, SSC, and LFS-SSC and smaller hip external rotation moments during conditions of SSC and LFS-SSC when compared to males. Therefore, although eccentric hip strength does not appear be a predictor of knee position, it may possibly be related to hip mechanics.

It is possible the relationship between hip strength and the control of knee abduction is stronger during jump landing tasks than during change-of-direction tasks. Intuitively, the relative need to control the decent of the center of mass is very different between the two tasks. Jump landings require the individual to come to a complete stop, whereas cutting tasks represent a momentary transition between two directions of travel. Hip abduction strength may thus serve a lesser role relative to other contributions to lower extremity dynamics in the control of knee abduction during cutting tasks. One limitation to these conclusions is related to the mode of hip strength measurement. We anticipated that eccentric hip strength would represent a functional measurement of proximal control associated with lower extremity alignment. The total excursion of the hip through abduction and adduction during the tasks may not represent the position in which peak eccentric hip strength was found to occur ($19.53 \pm 1.66^{\circ}$) in this study. It is possible that a relationship between hip strength and knee abduction exists when the angle at which the strength measurement is obtained replicates a hip position associated with peak frontal plane knee kinematics. Investigation of isometric hip strength may be warranted if the amount of hip abduction does not extend over a very large range during the dynamic tasks.

To our knowledge, this study was the first to compare standing Q-angle and eccentric hip strength measurements to frontal plane knee loading during running and change of direction tasks. Although these measurements are easy to administer clinically, the apparent lack of relationship to knee adduction moments suggests other factors may be important in predicting knee loading. It may be useful to consider coupled relationships of hip abductor/knee extensor, hip abductor/hip flexor, or hip extensor/knee extensor strength ratios. Inasmuch as hip and knee positions and loadings are inter-related, so too may be the strength of muscles associated with different joint and planar control during dynamic tasks. Dissimilarities between these strength ratios may better explain the different knee loading behaviors associated with activities related to ACL injury.

Although both the relationships between standing Q-angle, sex, and their interaction and between eccentric hip strength, sex, and their interaction were significant for the LFS condition, we were unable to determine which factor(s) satisfactorily contributed to the model. These results suggest that the performance of a lateral false step may be more sensitive to the combined effects of static lower extremity alignment, hip strength, and sex, but it is unclear which of these factors helps explain the observed differences. It is possible we lacked sufficient power to determine these relationships due to the size of the population included in this study.

Recently Chaudhari and Andriacchi[7] investigated the contribution of hip stiffness to ACL injury threshold in a dynamic model. In the conditions of neutral alignment, the injury threshold was significantly greater than when knee alignment changed to a valgus (abduction) position. Hip stiffness was also found to have a positive effect on injury threshold, such that an increase improved the threshold in both neutral and abducted alignments of the knee. Intuitively, increases in hip abduction strength should be related to the ability to increase hip stiffness during dynamic tasks. Although we did not directly measure hip stiffness in this study, our results indicate hip strength is not a predictor of knee abduction position or frontal plane knee loading during running and change of direction tasks. Dynamic hip abduction stiffness may be more related to anticipatory contraction of hip musculature during these tasks than overall hip abduction strength. Determining overall hip abductor activation may be more helpful in explaining these relationships and why ACL injury risk is higher for females than males during in vivo studies.

The results of this study do not support evidence of a relationship between static measurements of lower extremity alignment or eccentric hip abduction strength and frontal plane knee position and loading behaviors during running and rapid change of direction tasks. Standing Q-angle and eccentric hip abduction strength do not appear to differ in collegiate female and male basketball and soccer players. Hip abduction strength may be a better predictor of frontal plane knee kinematics during jump landing tasks than cutting tasks. Although female basketball and soccer players have been found to be at greater risk for ACL injuries than males when matched by age and sport, the present data do not help to explain this phenomenon.

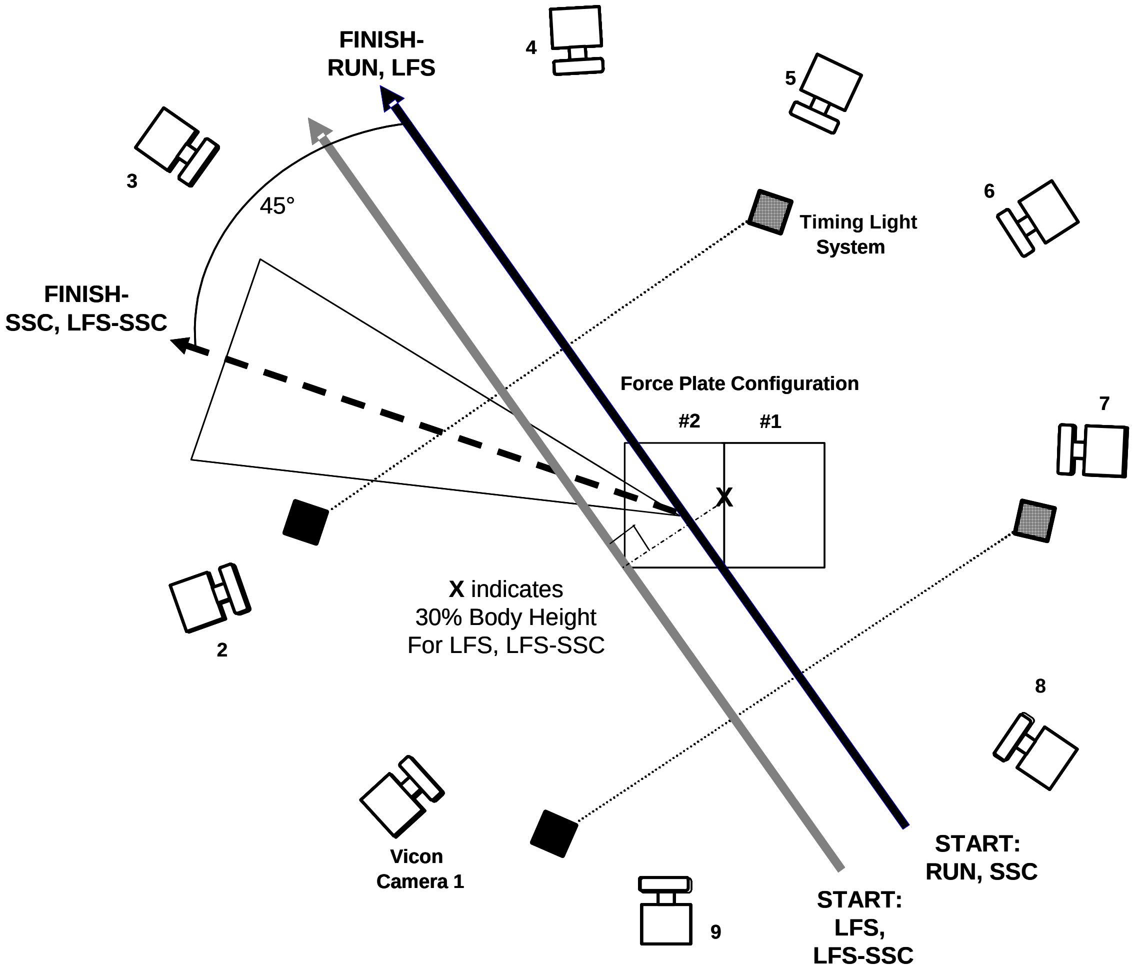


Figure 1. Laboratory set-up for data collection of running (RUN), lateral false step (LFS), sidestep cutting (SSC) and combination of lateral false step-sidestep cutting (LFS-SSC) maneuvers. Distance from the running lines for LFS and LFS-SSC maneuvers was determined to be 30% of body height and measured at a distance perpendicular to the direction of travel for both conditions. Change-of-direction angle for SSC and LFS-SSC conditions was set to 45 ± 10º. Tape was applied to the floor to indicate the different paths of travel for all conditions in addition to the acceptable range of 35-55º for the SSC and LFS-SSC conditions.

TABLE 1

Sex-based comparisons of standing Q-angle, hip abduction strength, and hip angle of peak torque development[a]

	Sex		
Variable	Females	Males	P-value
Standing Q-angle (º)	9.97 ± 3.83	7.60 ± 3.34	0.16
Eccentric hip abduction strength (Nm/(kg·m))	1.32 ± 0.18	1.25 ± 0.23	0.42
Angle of peak torque (º)[b]	19.65 ± 1.51	19.43 ± 1.87	0.78

[a] Values represent mean ± SD (Females 11, Males 10) for collegiate basketball and soccer athletes

[b] N = 20 (Females 10, Males 10)

TABLE 2

Relationship between standing Q-angle, sex, interaction of standing Q-angle and sex, with peak knee position and loading

Response Variable	Condition[c]	R	R^2	P-value
Knee Abduction Angle[a]	RUN	0.52	0.28	0.13
	LFS	0.63	0.40	0.03
	SSC	0.56	0.31	0.09
	LFS-SSC	0.38	0.15	0.43
Knee Adduction Moment[b]	RUN	0.59	0.09	0.67
	LFS	0.21	0.04	0.86
	SSC	0.28	0.03	0.70
	LFS-SSC	0.45	0.20	0.27

[a] Resulting R and R^2 values related to linear regression model (N = 21) with knee abduction angle (º) as the response variable and sex, standing Q-angle and the interaction between sex and SQA as explanatory variables.
[b] Resulting R and R^2 values related to linear regression model (N = 21) with knee adduction moment (% body weight x body height) as the response variable and sex, standing Q-angle and the interaction between sex and SQA as explanatory variables.
[c] RUN = running, LFS = lateral false step, SSC = sidestep cutting, LFS-SSC = combination of lateral false step and sidestep cutting.

TABLE 3
Relationship between eccentric hip strength, sex, interaction of hip strength and sex, with peak knee position and loading

Response Variable	Condition[c]	R	R^2	P-value
Knee Abduction Angle[a]	RUN	0.48	0.23	0.21
	LFS	0.68	0.46	0.01
	SSC	0.53	0.28	0.13
	LFS-SSC	0.38	0.15	0.43
Knee Adduction Moment[b]	RUN	0.37	0.14	0.45
	LFS	0.332	0.10	0.60
	SSC	0.50	0.24	0.18
	LFS-SSC	0.45	0.20	0.27

[a] Resulting R and R^2 values related to linear regression model (N = 21) with knee abduction angle (º) as the response variable and sex, eccentric hip strength (ECC), and the interaction between sex and ECC as explanatory variables.
[b] Resulting R and R^2 values related to linear regression model (N = 21) with knee adduction moment (% body weight* body height) as the response variable and sex, eccentric hip strength (ECC) and the interaction between sex and ECC as explanatory variables.
[c] RUN = running, LFS = lateral false step, SSC = sidestep cutting, LFS-SSC = combination of lateral false step and sidestep cutting.

REFERENCES

1. Agel J, Arendt EA, Bershadsky B. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med.* 2005;33(4):524-530.PMID: 15722283 Provider: OCLC

2. Andrews JR, McLeod WD, Ward T, Howard K. The cutting mechanism. *Am J Sports Med.* 1977;5(3):111-121.PMID: 860773 Provider: OCLC

3. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. *Am J Sports Med.* Nov-Dec 1995;23(6):694-701.8600737

4. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* Jun 2000;23(6):573-578.PMID: 10875418; Medline: 20331673 Provider: OCLC

5. Byl T, Cole JA, Livingston LA. What determines the magnitude of the Q angle? A preliminary study of selected skeletal and muscular measures. *J Sport Rehab.* 2000;9:26-34

6. Cahalan TD, Johnson ME, Liu S, Chao EY. Quantitative measurements of hip strength in different age groups. *Clin Orthop Relat Res.* 1989;246:136-145.PMID: 2766602 Provider: OCLC

7. Chaudhari AM, Andriacchi TP. The mechanical consequences of dynamic frontal plane limb alignment for non-contact ACL injury. *J Biomech.* 2006;39(2):330-338.PMID: 16321635 Provider: OCLC

8. Claiborne TL, Armstrong CW, Gandhi V, Pincivero DM. Relationship between hip and knee strength and knee valgus during a single leg squat. *J Appl Biomech.* Feb 2006;22(1):41-50.PMID: 16760566 Provider: OCLC

9. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech.* 1996;29(9):1223-1230.PMID: 8872282 Provider: OCLC

10. Delp SL, Hess WE, Hungerford DS, Jones LC. Variation of rotation moment arms with hip flexion. *J Biomech.* 1999;32(5):493-501.PMID: 10327003 Provider: OCLC

11. Frank CB, Jackson DW. The science of reconstruction of the anterior cruciate ligament. *J Bone Joint Surg Am.* 1997;79(10):1556-1576.PMID: 9378743 Provider: OCLC

12. Fredericson M, Cookingham CL, Chaudhari AM, Dowdell BC, Oestreicher N, Sahrmann SA. Hip abductor weakness in distance runners with iliotibial band syndrome. *Clin J Sport Med.* Jul 2000;10(3):169-175.PMID: 10959926 Provider: OCLC

13. Golden GM. *The effect of altering the width of a single step of a running stride on knee joint kinematics and kinetics* [In Lieu of Thesis, unpublished manuscript]. Corvallis: Nutrition and Exercise Sciences, Oregon State University; 2005.

14. Golden GM, Pavol MJ, Hoffman MA. *Sex differences in kinematics and kinetics of the knee and hip during running and rapid change of direction tasks* [Dissertation]. Corvallis: Nutrition and Exercise Sciences, Oregon State University; 2007.

15. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, Garrick JG, Hewett TE, Huston L, Ireland ML, Johnson RJ, Kibler WB, Lephart S, Lewis JL, Lindenfeld TN, Mandelbaum BR, Marchak P, Teitz CC, Wojtys EM. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *The Journal of the American Academy of Orthopaedic Surgeons.* May-Jun 2000;8(3):141-150.PMID: 10874221; Medline: 20334910 Provider: OCLC

16. Griffin LY, Albohm MJ, Arendt EA, Bahr R, Beynnon BD, Demaio M, Dick RW, Engebretsen L, Garrett WE, Jr., Hannafin JA, Hewett TE, Huston LJ, Ireland ML, Johnson RJ, Lephart S, Mandelbaum BR, Mann BJ, Marks PH, Marshall SW, Myklebust G, Noyes FR, Powers C, Shields C, Jr., Shultz SJ, Silvers H, Slauterbeck J, Taylor DC, Teitz CC, Wojtys EM, Yu B. Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *American Journal of Sports Medicine.* 2006;34:1512-1532.PMID: 16905673 Provider: OCLC

17. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng.* 1983;105(2):136-144.PMID: 6865355 Provider: OCLC

18. Guerra JP, Arnold MJ, Gajdosik RL. Q angle: effects of isometric quadriceps contraction and body position. *J Orthop Sports Phys Ther.* Apr 1994;19(4):200-204.PMID: 8173567 Provider: OCLC
19. Hollman JH, Kolbeck KE, Hitchcock JL, Koverman JW, Krause DA. Correlations between hip strength and static foot and knee posture. *J Sport Rehab.* 2006;15:12-23
20. Horton MG, Hall Tl. Quadriceps femoris muscle angle: normal values and relationships with gender and selected skeletal measures. *Phys Ther.* 1989;69:897-901.PMID: 2813517 Provider: OCLC
21. Huston LJ, Greenfield ML, Wojtys EM. Anterior cruciate ligament injuries in the female athlete. Potential risk factors. *Clin Orthop Relat Res.* 2000;372:50-63.PMID: 10738414 Provider: OCLC
22. Ireland ML. The female ACL: why is it more prone to injury? *The Orthopedic clinics of North America.* Oct 2002;33(4):637-651.PMID: 12528906; Medline: 22416240 Provider: OCLC
23. Jacobs C, Mattacola C. Sex differences in eccentric hip-abductor strength and knee-joint kinematics when landing from a jump. *J Sport Rehabil.* 2005;14:346-355
24. Jacobs CA, Uhl TL, Mattacola CG, Shapiro R, Rayens WS. Hip abductor function and lower extremity landing kinemtaics: sex differences. *Journal of Athletic Training.* 2007;42:76-83
25. Kirkendall DT, Garrett WE. The anterior cruciate ligament enigma. Injury mechanisms and prevention. *Clinical orthopaedics and related research.* 2000;372:64-68.PMID: 10738415 Provider: OCLC
26. Lathinghouse LH, Trimble MH. Effects of isometric quadriceps activation on the Q-angle in women before and after quadriceps exercise. *J Orthop Sports Phys Ther.* Apr 2000;30(4):211-216.PMID: 10778798 Provider: OCLC
27. Leetun DT, Ireland ML, Willson JD, Ballantyne BT, Davis IM. Core stability measures as risk factors for lower extremity injury in athletes. *Med Sci Sports Exerc.* Jun 2004;36(6):926-934.PMID: 15179160 Provider: OCLC
28. Maletius W, Messner K. Eighteen- to twenty-four-year follow-up after complete rupture of the anterior cruciate ligament. *Am J Sports Med.* 1999;27(6):711-717.PMID: 10569355 Provider: OCLC
29. Mansour JM, Pereira JM. Quantitative functional anatomy of the lower limb with application to human gait. *J Biomech.* 1987;20(1):51-58.PMID: 3558428 Provider: OCLC
30. McLean SG, Myers PT, Neal RJ, Walters MR. A quantitative analysis of knee joint kinematics during the sidestep cutting maneuver. Implications for non-contact anterior cruciate ligament injury. *Bull Hosp Jt Dis.* 1998;57(1):30-38.PMID: 9553700 Provider: OCLC
31. McNair PJ, Marshall RN, Matheson JA. Important features associated with acute anterior cruciate ligament injury. *N Z Med J.* 1990;103:537-539.PMID: 2243642 Provider: OCLC
32. Messier SP, Davis SE, Curl WW, Lowery RB, Pack RJ. Etiologic factors associated with patellofemoral pain in runners. *Med Sci Sports Exerc.* Sep 1991;23(9):1008-1015.PMID: 1943620 Provider: OCLC
33. Moul J. Differences in selected predictors of anterior cruciate ligament tears between male and female NCAA Division I collegiate basketball players. *Journal of Athletic Training.* 1998/04//1998 Apr-Jun 1998;33(2):118-121
34. Pantano KJ, White SC, Gilchrist LA, Leddy J. Differences in peak knee valgus angles between individuals with high and low Q-angles during a single limb squat. *Clin Biomech (Bristol, Avon).* Nov 2005;20(9):966-972.PMID: 16051403 Provider: OCLC
35. Pollard CD, Davis IM, Hamill J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech.* 2004;19(10):1022-1031.PMID: 15531052 Provider: OCLC
36. Powers CM. The Influence of Altered Lower-Extremity Kinematics on Patellofemoral Joint Dysfunction: A Theoretical Perspective. *J Orthop Sports Phys Ther.* 2003;33:693-646
37. Romani W, Patrie J, Curl LA, Flaws JA. The correlations between estradiol, estrone, estriol, progesterone, and sex hormone-binding globulin and anterior cruciate ligament stiffness in healthy, active females. *J Womens Health.* 2003;12(3):287-298.PMID: 12804359 Provider: OCLC
38. Shambaugh JP, Klein A, Herbert JH. Structural measures as predictors of injury basketball players. *Med Sci Sports Exerc.* May 1991;23(5):522-527.1906569

39. Shultz SJ, Kirk SE, Johnson ML, Sander TC, Perrin DH. Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Med Sci Sports Exerc.* 2004;36(7):1165-1174.PMID: 15235320 Provider: OCLC

40. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. *Clin Biomech.* 2006;21:41-48.PMID: 16209900 Provider: OCLC

41. Vaughan CL, Davis, B. L., O'Conner, J. C. *Dynamics of Human Gait*. 2nd ed. Cape Town, South Africa: Kiboho Publishers; 1999.

42. Willson JD, Ireland ML, Davis I. Core strength and lower extremity alignment during single leg squats. *Med Sci Sports Exerc.* May 2006;38(5):945-952.PMID: 16672849 Provider: OCLC

43. Winter DA. *Biomechanics and motor control of human movement*. Second ed. Toronto: Wiley-Interscience; 1990.

44. Woodland LH, Francis RS. Parameters and comparisons of the quadriceps angle of college-aged men and women in the supine and standing positions. *Am J Sports Med.* Mar-Apr 1992;20(2):208-211.PMID: 1558251 Provider: OCLC

Patterns of lower extremity kinematic and kinetic measures for females and males during running and rapid change-of-direction tasks

Grace M. Golden, Mark A. Hoffman, Michael J. Pavol

INTRODUCTION

Documented incidence of anterior cruciate ligament (ACL) rupture and the sex bias associated with this injury in collegiate basketball and soccer players has remained consistent over the last decade.[1, 3] The short-term and long-term complications associated with ACL rupture are well documented and range from time loss from activity to a high risk for developing knee osteoarthritis and degeneration.[14] There is apparent high agreement between the clinically observed mechanism of ACL injury and in situ biomechanical investigations of ACL loading patterns.[16, 17, 31] Research has shown that movements that combine a relatively extended knee (0-30º) which is concomitantly internally rotated and abducted are typical of the ACL injury mechanism.[4, 13, 23] One movement behavior associated with the production of this high risk position is a sudden change in direction while running, i.e. sidestep cutting.[18] Although knee abduction has been suspected to contribute to ACL injury during jump landing for females[12], there is less consensus regarding sex differences and predictors of injury for sidestep cutting.[7, 21, 22, 24, 29] Therefore, is it necessary is to continue to advance our understanding of the mechanism of ACL injury and potential precipitating factors in the athletic population, endeavoring to help prevent this injury during rapid change-of-direction tasks.

Biomechanical studies of sidestep cutting have primarily focused on quantifying knee kinematics and kinetics, however, there has been recent consideration of the role of hip position and loading patterns. There is lack of agreement as to whether sex differences exist during the performance of sidestep cutting (SSC) tasks. Although females have been found to demonstrate greater positions of knee abduction[15, 20] and less internal rotation[21] and knee flexion[15, 21] compared to males in SSC tasks, other studies with similar experimental design have not established a sex difference for knee kinematics for SSC tasks.[24, 29] Similarly, some researchers have reported females generate greater knee abduction[19, 20, 29] and smaller internal rotation[19] moments compared to males during SSC tasks while others[24] have not detected differences in knee kinets during SSC tasks.

Consideration of the role of hip kinematics and kinetics is equally divided. McLean et al.[21] postulated greater hip external rotation is concomitant with greater knee abduction. However, Pollard et al.[26] reported that, although females exhibit less peak hip abduction than males, there is no effect of sex on hip rotation or knee kinematics. Additionally, these same authors found no differences in hip loading patterns between males and females in the frontal and transverse planes. Differences in frontal plane hip kinematics without dissimilarity in knee kinematics provides disconnect between the expected relationships of the proximal and distal joints of the lower extremity. It is quite possible that the inability to conclusively explain sex differences from both kinematic and kinetic perspectives, is due to the investigation of these variables in isolation. It is well established that during rapid changes in direction, many of the traditional variables are highly inter-related. Therefore we utilized statistical techniques to combine several of the related variables into factors that describe the principal components of the movement behavior.

Rapid changes of direction represent highly inter-related multi-joint coordinative movements. The potential danger in looking at variables in isolation stems from the inability to compare how one variable is related to, or affects, another. Therefore, considering combined patterns of joint motion and loading across two joints may be useful in describing the movement behavior associated with rapid change-of-direction tasks and may better delineate the presence of sex differences. In fact, reducing high-dimensional data of several kinematic and kinetic variables to a smaller number of descriptive patterns that capture the movement behavior has been encouraged for biomechanical studies.[6, 10] This approach may contribute to understanding of how kinematic and kinetic knee and hip variables are related, improving the ability to determine how rapid change-of-direction tasks are associated with the ACL injury mechanism.

This study was designed to systematically investigate patterns of knee and hip kinematics and kinetics that emerged during running and three variations of rapid change of direction tasks of increasing complexity and compare them across sexes and change-of-direction tasks. The primary purpose of this study was to identify a set of factors which characterize knee and hip position and loading behaviors in collegiate athletes during conditions of normal running and rapid change-of-direction tasks. The secondary purpose of this study was to determine if there are differences in the patterns of movement displayed by females and males. Based upon these objectives, we hypothesized that females would exhibit to a great extent those lower extremity patterns that place the ACL at greater risk of injury, as would variations of the sidestep cutting maneuver.

METHODS

Twenty-one subjects (11 females: 19.8 ± 1.5 yrs, 175.8 ± 6.3 cm, 69.9 ± 6.6 kg; 10 males: 20.7 ± 2.5 yrs, 188.6 ± 15.9 cm, 89.8 ± 21.7 kg) , basketball and soccer athletes from four college and university programs, were enrolled in the study. During initial screening, subjects completed demographic and health history questionnaires. Exclusion criteria included a past history of any reconstructive surgical procedure for the treatment of a lower extremity injury within the past twelve month period. All subjects were required to have an intact ACL. They could not be currently receiving supervised medical care for an injury to the upper or lower extremity preventing their participation in regular conditioning or practice sessions nor could they have suffered an injury to the lower extremity resulting in time loss from participation exceeding three weeks within the past six months. Additionally, female subjects were screened for a regular menstrual cycle (28-32 days[27, 28]) for the past 3 months and participated in the testing session within 1-9 days of the start of menses (6.5 ± 1.8 days). All participants were of equivalent sport experience (females: 8.45 ± 1.97 yrs, males: 7.00 ± 1.76 yrs; $p > 0.05$) and participating in either off-season conditioning (basketball) or a non-traditional competitive season (soccer). Recruitment and participation of subjects was approved by a University Institutional Review Board for the Protection of Human Subjects and all subjects provided informed consent prior to participation.

Kinematic and kinetic data acquisition were performed with a Vicon 612 optical motion capture system utilizing 9 high-resolution MCam2 digital cameras (Vicon, Lake Forest, CA), interfaced with two Bertec Model 4060-08 force plates (Bertec, Columbus, OH) that were configured to function as a single force plate. Sixteen markers (9mm diameter) were affixed bilaterally on the lower extremity with double sided tape in order to identify body segment positions during static and dynamic trials. A customized marker set, based upon the work of Vaughan et al.[30] was utilized for this study (5th metatarsal, heel, ankle, tibia, knee, thigh, anterior superior iliac spine, and posterior superior iliac spine). Kinematic data were collected at 120Hz and kinetic data were collected throughout foot contact at 1080 Hz.

Figure 1 illustrates the laboratory set up for the running and change-of-direction tasks. In order to identify the path of travel for RUN and SSC conditions, yellow tape was affixed to the floor in a line crossing the center of the force plate. A second, parallel, blue tape line was placed on the floor, crossing the left corner of the force plate to indicate the path of travel for the LFS and LFS-SSC conditions. A divergent tape line, originating from the force plates, was placed at an angle of 45º to the left of the other lines to identify the pathway for subjects to successfully perform the SSC and LFS-SSC change-of-direction maneuvers. Colored adhesive tape was applied on the force plate to indicate a distance of 30% of body height for LFS and LFS-SSC conditions. This distance was measured perpendicularly to the right from the center point of the tape line crossing the left corner of the force plate. Cameras were positioned so each marker was visible by multiple cameras when the right foot was in contact with the force plate. Each trial consisted of running a distance of approximately 10m from start to finish.

On the day of testing, subjects reported to the testing laboratory. Height (cm) was measured (with shoes on) with a wall mounted stadiometer. Weight (lbs) was recorded with a floor scale and later converted to kg. Next, subjects performed a 5 minute warm-up by running on a treadmill (2.5-3.1 $m \cdot s^{-1}$). During this time, adhesive tape was applied to the force platform at a location indicating a distance equivalent to 30% of body height from the line of travel for the LFS and LFS-SSC conditions.

After the warm-up period, subjects were allowed 3-5 practice trials for each of the running and change-of-direction tasks in the order of RUN, LFS, SSC, LFS-SSC. These practice trials enabled researchers to determine the most effective starting location so that the running stride prior to force plate contact was fluid and consistent, and improved subject familiarization with the specific tasks. Throughout the practice and testing trials, the speed of the subject was monitored through the use of timing lights placed 1.5 m to the rear and front of the force plates, at a total distance of 3 m apart. Trials were considered acceptable when speed equated to 3.5 $m \cdot s^{-1}$ ± 10% (3.15 - 3.85 $m \cdot s^{-1}$). Recovery was accommodated by having subjects walk back to the starting location between each trial. Verbal instructions for running speed and technique were standardized for all subjects. Subjects were directed to perform all trials without a reduction of speed into the RUN, LFS, SSC, and LFS-SSC maneuvers, discouraging leaping or hesitating prior to force plate contact. Testing trials were discarded if the subject made foot contact on the force plates with the left foot, failed to make contact with the force plates with the entire right foot, missed the LFS and LFS-SSC tape mark on the force plate, initiated the LFS or LFS-SSC maneuver by crossing the left foot to the right of the directional guide placed on the floor, or completed the SSC or LFS-SSC cut outside a range of 45º ± 10º. Colored tape was placed on the floor to indicate the boundaries of 35-55º.

Following familiarization of running and change-of-direction tasks, markers were affixed to the subjects. Each subject wore form-fitting spandex shorts of a dark color and a pair of their own team-issued running shoes for data acquisition trials. Foot markers were placed on the outside of the running shoe. A static trial was collected for the calibration of segmental orientation, with the subject positioned in anatomical neutral (arms at 90º abduction) while standing on the force plates. Ten successful trials were recorded for each of the four conditions in the order of: RUN, LFS, SSC, and LFS-SSC. During each trial, kinematic and kinetic data were measured for the subject's right limb throughout contact with the force plate during stance phase. Subsequent to the completion of all running and change-of-direction trials, foot length (shoe length) and pelvis, knee, and ankle width were measured with a joint anthropometer (Lafayette Instrument Company, Lafayette, IN). Three measurements were obtained for each measurement by the same researcher and averaged for data analysis. All data were collected in a single testing session.

Three degrees of freedom were assigned to the major joints of interest, with the corresponding rotations of flexion-extension, abduction-adduction, and internal-external rotation for the hip, knee, and ankle joints. Raw data for all dynamic trials were used to determine the three-dimensional paths of the markers and cubic spline interpolation was performed to fill small gaps (less then 50ms) with Vicon Workstation software. Kinematic data were filtered using a fourth-order zero-lag Butterworth low-pass filter with a cut-off frequency of 16Hz, as determined through a residual analysis.[32] Marker positions from the static, standing trial and anthropometric measures were used to establish a set of transformations for determining joint center locations and body segment orientations. These transformations were used to determine the paths of the joint centers and the orientations of the body segments over the entire stance phase of running. From these, the joint angles (degrees) at the stance knee and hip were determined by the convention of Grood and Suntay[11]. Knee internal rotation angles during the static trial were defined to be zero. The computed kinematics were also combined with the force plate data to determine the internal joint moments acting at the knee and hip during stance using 3-dimensional inverse dynamics methods. Foot progression angle was determined as the foot abduction angle at midstance, relative to the direction of the original path of travel. Segmental inertial parameters were determined as outlined by de Leva for female and male data.[8] Joint moments were expressed with respect to the assumed joint axes of rotation. Joint moments (Nmm) were normalized to body height (mm) and body weight (N) and reported as a percentage of body height (bh) and body weight (bw). A customized Matlab program (Student version 7, Mathworks, Natick, MA) was created to extract peak joint angles and moments from initial contact to 30º of knee flexion during stance for all dynamic trials. An average of the peak values from the ten trials of each condition by each subject was used for statistical analysis.

Analysis was conducted on 18 measures of lower extremity position and loading across the four conditions of running and change-of-direction tasks, including peak knee angles of extension, abduction, internal rotation; peak hip angles of flexion, abduction, external rotation; foot progression angle; peak knee moments of flexion, extension, abduction, adduction, internal rotation, external rotation; and peak hip moments of extension, abduction, adduction, internal rotation, external rotation. Pearson correlations were computed between the measures of foot progression, knee and hip position, and loading variables. All measures were found to be correlated and included in a factor analysis. Factors were extracted using principal components analysis and the number of factors retained in the model selected based upon the results of a scree plot. A direct oblimin factor rotation with Kaiser normalization was employed. Factor scores were computed using the regression method.

Four separate 2 way repeated measures ANOVA (Sex;2 X Task;4) were conducted to compare the factor scores for females and males across the conditions of RUN, LFS, SSC, and LFS-SSC. Post hoc comparisons between the four conditions were conducted utilizing the Tukey's HSD test. The level of significance was set at $\alpha = 0.05$. Analyses were performed using SPSS 13.0 (Chicago, IL).

RESULTS

Four common factors were identified as underlying the eighteen measures of lower extremity kinematics and kinetic (Table 1). These four factors explained 73.05% of the variance in the kinematic and kinetic measures.

Factor one, *Knee Extension*, was associated with the greatest number of lower extremity measures. Through this factor, a knee which is more extended during the early portion of stance was related to lesser peak knee abduction and greater peak moments in knee abduction, extension, external rotation, hip abduction and hip internal rotation. Factor two, *Hip Abduction*, related a lower extremity pattern of greater peak hip abduction position with a greater peak knee adduction moment, greater peak hip adduction moment, and a greater peak hip external rotation moment. Factor three, *Hip Flexion*, related a position of greater hip flexion with a greater peak knee flexion moment and peak hip extension moment. Lastly, factor four, *Foot Abduction,* related a position of greater foot abduction (toe-out) with less peak knee internal rotation, greater peak hip external rotation, and a lesser peak knee internal rotation moment.

A significant main effect of condition was evident for *Knee Extension* ($p < 0.001$). Post hoc analysis showed the presence of *Knee Extension* was lesser for LFS-SSC than for both the RUN and LFS conditions ($p < 0.05$; Table 2). There was not a significant main affect of sex ($p = 0.95$) or an interaction between sex and condition ($p = 0.48$) for *Knee Extension*.

Similarly, a significant main effect of condition was evident for *Hip Abduction* ($p < .001$). Post hoc analysis determined that the presence of *Hip Abduction* was greater for LFS, SSC and LFS-SSC conditions than for RUN ($p < 0.05$; Table 2). Additionally, the presence of *Hip Abduction* was greater for the LFS-SSC condition than for LFS and SSC ($p < 0.05$). There was not a significant main effect of sex ($p = 0.10$) or an interaction between sex and condition ($p = 0.63$) for *Hip Abduction.*

A significant main effect of condition was also evident for *Hip Flexion* ($p = 0.003$). However, post hoc analysis showed no significant difference between any of the four conditions ($p > 0.05$, Table 2). There was not a main effect of sex ($p = 0.09$) or interaction between sex and condition ($p = 0.17$) for *Hip Flexion*.

Lastly, a significant main effect of condition was evident for *Foot Abduction* ($p < 0.001$). Post hoc analysis determined that the presence of *Foot Abduction* was lesser for the condition of SSC than for RUN and lesser for the conditions of SSC and LFS-SSC than for LFS ($p < 0.05$; Table 2). There was not a main effect of sex ($p = 0.80$) or interaction between sex and condition ($p = 0.64$) for *Foot Abduction*.

DISCUSSION

The main objective of this study was to identify patterns that characterize knee and hip position and loading behaviors during running and change-of-direction tasks and to determine that extent to which these patterns differs between tasks. Four patterns, defined as factors, were identified as primary contributors to the expression of kinematic and kinetic behaviors during the dynamic tasks, explaining 73.05% of the variance. These patterns provide evidence of common relationships between lower extremity kinematics and kinetics during dynamic tasks.

The first factor, *Knee Extension,* represented a pattern in which greater knee extension was related to lesser knee abduction, greater extension, abduction, and external rotation moments at the knee, and greater abduction and internal rotation moments at the hip in the early part of stance. This factor identifies a relationship between variables seen during the initial loading phase of running.[9] During early stance in running, the knee is relatively extended with less abduction. Knee external rotation moments and hip internal rotation moments were coupled, implying that internal forces functioned to simultaneously resist internal knee rotation and external hip rotation. The association between this pattern and running mechanics is further supported because *Knee Extension* is more strongly represented during running and LFS when compared to the LFS-SSC task. We would expect an increase in the knee extension moment to be hazardous, as loading through the sagittal plane with a larger anterior tibial force increases loading of the ACL.[17] However, this factor couples larger knee extension moments with greater knee abduction and external rotation moments. This combined frontal and transverse plane loading pattern has not been related to loading of the ACL.[17] The strong presence of this pattern in running further supports the notion that this factor represents a pattern that is not hazardous to the ACL, as injuries do not commonly occur during running tasks. The presence of this pattern in the LFS maneuver also suggests this task may not place an individual at increased risk for injury through a sagittal plane mechanism.

Factors two and three (*Hip Abduction*, *Hip Flexion*) expressed patterns associated with hip position. *Hip Abduction* represented a pattern in which greater hip abduction was associated with greater knee adduction moments, hip adduction moments, and hip external rotation moments. We would expect positions of greater hip abduction to be associated with greater hip adduction moments. However, this pattern also demonstrated that an abducted hip was coupled with larger knee adduction moments. This would suggest increases in hip abduction may simultaneously increase potential ACL injury risk since internal moments of knee adduction are believed to load the ACL. This coupling of hip position and knee loading may serve to control knee abduction during the tasks. Other researchers have recognized this relationship and suggest that increased hip abduction contributes to a greater laterally directed ground reaction force, in turn producing larger knee adduction moments.[29] The presence of greater external rotation moments in this pattern suggested a need for more transverse plane control during tasks in which this pattern is strongly represented. It is reasonable that *Hip Abduction* was represented to a greater degree during the LFS, SSC and LFS-SSC conditions when compared to running. One of the primary components of two of these tasks required the athletes to increase step width in a lateral direction which would increase hip abduction. Additionally, athletes may have greater hip abduction as a direct result of preparing for the change of direction associated with the task. The coupling of these variables demonstrates how this pattern, which includes greater hip abduction, may place an individual at greater risk for ACL injury. Frontal plane motion of the knee, specifically knee abduction, is believed to increase ACL loading.[16, 17] Knee adduction moments are thought to control knee abduction during change-of-direction tasks in which increased ACL loading may occur. [20, 29] Combination of a laterally directed step with a sidestep cut appears to be more hazardous since the *Hip Abduction* pattern is most strongly represented during the LFS-SSC task.

Coupling of knee and hip kinetic and kinematic measures in *Hip Flexion* related increased hip flexion with larger peak knee flexion moments and hip extension moments. It stands to reason that increased hip flexion would be related to greater hip extensor moments, as greater hip flexion would increase the demand on the hip extensors. Association of greater hip flexion with a larger knee flexion moment is likely related to involvement of the hamstring muscle group, which acts to flex the knee and secondarily extend the hip. The coupling of these hip and knee variables suggest the more the hip is flexed, the greater the associated knee flexion moment. Pollard et al.[25] suggest the demonstration of decreased hip flexion during a change-of-direction task may result in the need for an alternative strategy for controlling deceleration in the sagittal plane. These authors propose that sagittal plane loading at the hip and knee, in the presence of adequate hip flexion, may be protective of the ACL. The coupling of variables associated with *Hip Flexion* in this investigation supports this concept. This pattern provides evidence of the role of proximal control in positively influencing loading at the knee. *Hip Flexion* was found to be equally represented across all four dynamic tasks, indicating that athletes in this study performed each of the tasks with similar hip flexion and hip extension and knee flexion moments through the sagittal plane.

Factor four, *Foot Abduction,* primarily associated knee and hip measures in the transverse plane and was the only factor to include foot abduction. A pattern of increased foot abduction (increased externally rotated foot progression angle) was related to lesser knee internal rotation and greater hip external rotation in addition to a reduction in knee internal rotation moment. Patterning of foot, knee and hip motion through the transverse plane indicates the lower extremity was relatively externally rotated during early stance. It stands to reason that knee internal rotation moments would increase in this situation; however the reduction of this moment suggests other factors contribute to the control of knee external rotation. Conceivably, position of the lower extremity is organized to reduce loading of the knee through the transverse plane. In particular, external hip rotation is strongly associated with this pattern. McLean et al.[21] suggest that greater external rotation of the hip will lead to increased knee abduction and therefore greater possibility of ACL loading. The coupling of hip external rotation and foot abduction in this pattern without strong association with knee abduction may indicate the proposed relationship is not as direct as previously recommended.

Researchers have found an increase in foot abduction results in reduced hip external rotation moments[5] and knee adduction moments (external moments)[2] during walking. It may be the lack of a similar association between foot abduction and hip and knee loading found in *Foot Abduction* is due to the differences in the nature of the dynamic tasks that were investigated. Since ACL loading has been shown to be lower during external knee rotation compared to internal rotation[17], we would expect this pattern to afford protection to the ACL. Presence of *Foot Abduction* was greater in the LFS compared to SSC and LFS-SSC conditions and greater in running compared to the SSC. The overall reduction of internal rotation of the lower extremity in running and the LFS tasks may afford relative task specific protection of the ACL. In contrast, the SSC task was found to have the smallest amount of this pattern and may be one reason sidestep cutting is associated with ACL injury. In effect, the role of loading through the transverse plane may contribute more to the mechanism of ACL injury during this task when compared to other change-of-direction tasks.

The second objective of this study was to determine whether there were differences between females and males in the expression of the different factors. Contrary to expectations, there was not an effect of sex on any of the four factors for any of the four conditions of running and rapid change-of-direction tasks. This may be related to relative homogeneity of the subject population, as all were collegiate athletes with equivalent years of sport experience, despite different sport specializations (basketball and soccer). Additionally, the size of the subject pool may not have been large enough, limiting the ability to detect differences across sex.

The use of a principal components analysis (PCA) can be a powerful tool in evaluating relationships in inter-related multi-joint coordinative movements, helping to extract the most relevant information.[6] However, it is not without limitations. PCA has relative scaling sensitivity, meaning summarizing a set of variables with different units may affect the attention given to the variable in the analysis. Although PCA allows the researcher to gain insight into the underlying factors associated with a behavior, it does not directly measure the behavior. Rather, the researcher considers which variables are strongly influenced by the factor. In the case of this study, we were able to consider combined patterns of joint motion and loading across two joints, but not directly measure the underlying value of the variable of interest.

In summary, the results of this study indicate that four key factors are associated with the coupling of knee and hip position and loading patterns during running and change-of-direction tasks. Running was associated with a pattern in which sagittal, frontal, and transverse plane loading of the lower extremity appeared to be protective of the ACL. Increasing hip abduction concomitantly raises frontal plane loading of the knee in a manner associated with increased knee abduction, which is believed to load the ACL. Stepping laterally prior to a sidestep cut will tend to increase the presence of this loading pattern and may be one reason athletes are susceptible to ACL injury during this type of rapid change-of-direction task. Increasing hip flexion enhanced sagittal plane loading of the hip and knee such that the ACL would be protected. We might expect if athletes perform change-of-direction tasks in a more upright position, they will experience a reduction in the beneficial coupling of hip extension and knee flexion moments, increasing susceptibility to ACL injury. Relative external rotation of the lower extremity may afford protection of the ACL when frontal and sagittal plane loading is held constant. Although the lateral step is associated with increased knee loading through the frontal plane, the associated reduction of transverse plane loading may lessen this detrimental effect. Sidestep cutting was performed with relative internal rotation of the lower extremity, which may help to explain how this change of direction task can lead to ACL injury.

From a clinical perspective, it is apparent that knee and hip position and loading behaviors are highly inter-related. The manner in which athletes rapidly change direction may be linked to different mechanisms associated with ACL loading. The results of this study imply that it is difficult to attribute a single parameter as predictive of ACL injury during these complex tasks. Females and males who adopt protective coupled foot, knee, and hip behavior will likely be at less risk of ACL injury. Further study is necessary to identify what factors will tend to reduce these positive relationships and increase the risk of ACL injury.

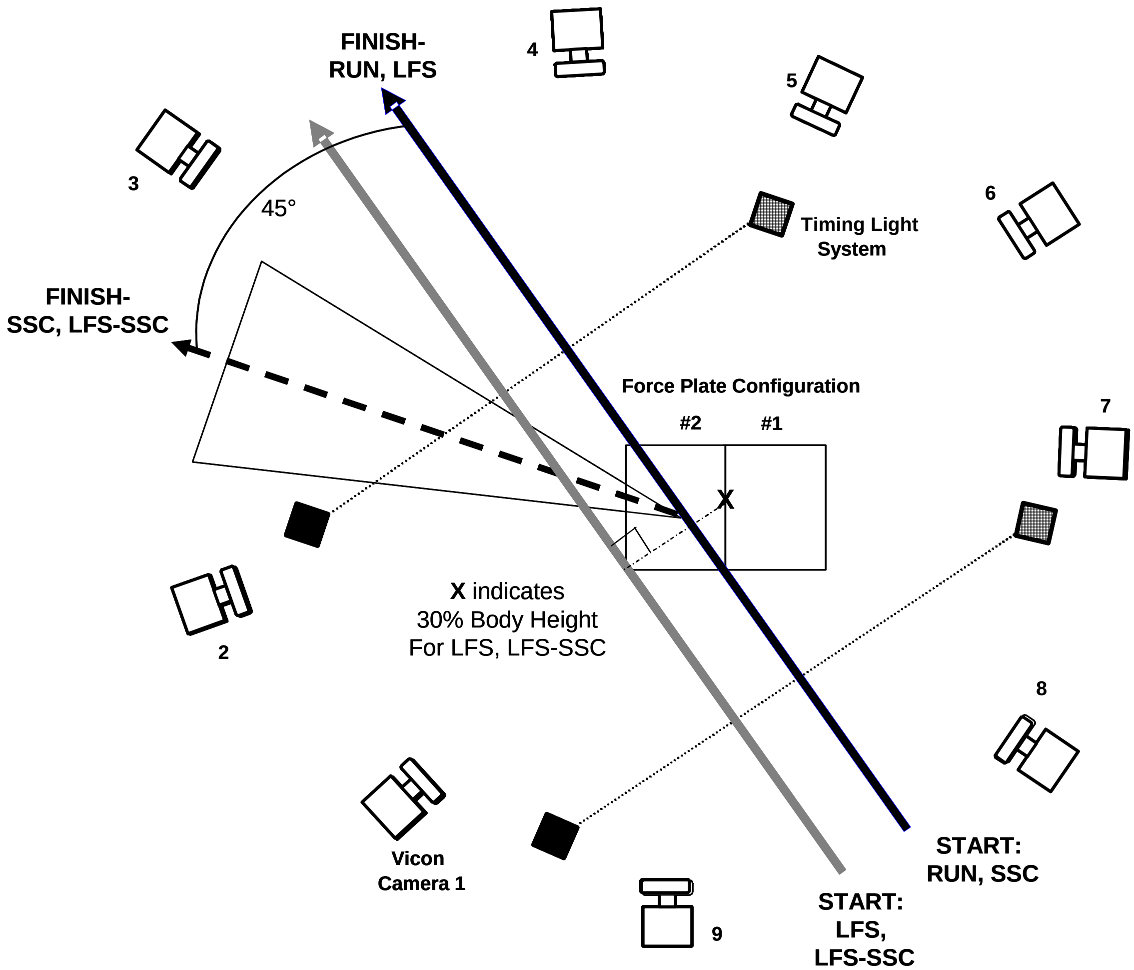


Figure 1. Laboratory set-up for data collection of running (RUN), lateral false step (LFS), sidestep cutting (SSC) and combination of lateral false step-sidestep cutting (LFS-SSC) maneuvers. Distance from the running lines for LFS and LFS-SSC maneuvers was determined to be 30% of body height and measured at a distance perpendicular to the direction of travel for both conditions. Change-of-direction angle for SSC and LFS-SSC conditions was set to 45 ± 10º. Tape was applied to the floor to indicate the different paths of travel for all conditions in addition to the acceptable range of 35-55º for the SSC and LFS-SSC conditions.

TABLE 1
Pattern matrix for four factors describing lower extremity kinematics and kinetics during running and change-of-direction tasks[a]

	Factor				
	Knee Extension	Hip Abduction	Hip Flexion	Foot Abduction	r^2
Kinematic Measures					
Knee extension	0.72				0.73
Knee abduction	-0.51				0.46
Knee internal rotation				-0.60	0.67
Hip flexion			0.74		0.68
Hip abduction		0.80			0.67
Hip external rotation				0.86	0.74
Foot progression angle				0.78	0.62
Kinetic Measures					
Knee flexion			0.63		0.68
Knee extension	0.75				0.63
Knee abduction	0.90				0.90
Knee adduction		0.90			0.95
Knee internal rotation				-0.62	0.51
Knee external rotation	0.69				0.70
Hip extension			0.81		0.84
Hip abduction	0.91				0.85
Hip adduction		0.87			0.90
Hip internal rotation	0.82				0.76
Hip external rotation		0.81			0.85

[a]Values represent the weighting (-1 to 1) of each factor in a linear combination describing each measure across all conditions of run and change of direction tasks. Large weights indicate a factor and kinematic or kinetic measure are strongly related. Weights with magnitudes less than 0.50 are not shown. The communality (r^2) indicates the proportion of variance explained by the four factors.

Table 2
Comparison of kinematic and kinetic factor scores across running and change-of-direction tasks[a]

Factor	Condition[b] RUN	LFS	SSC	LFS-SSC	P-value
Knee extension	0.30 ± 0.74	0.24 ± 0.95	-0.13 ± 1.10	-0.42 ± 1.10[cd]	< 0.001
Hip abduction	-1.10 ± 0.29	-0.21 ± 0.37[c]	0.02 ± 0.53[c]	1.29 ± 0.78[cde]	< 0.001
Hip flexion	0.13 ± 0.42	-0.43 ± 0.89	0.36 ± 0.93	-0.14 ± 1.40	0.003
Foot abduction	0.30 ± 0.71	0.53 ± 0.82	-0.55 ± 1.01[cd]	-0.29 ± 1.08[d]	< 0.001

[a]Factor scores each have a mean of 0.0 and standard deviation of 1.0 across subjects (i.e., scores are in units of standard deviations from the population average; N = 21).
[b]RUN = running; LFS = lateral false step; SSC = sidestep cutting; LFS-SSC = combination of lateral false step and sidestep cutting.
[c]significantly different than RUN (Tukey's HSD $p < 0.05$)
[d]significantly different than LFS (Tukey's HSD $p < 0.05$)
[e]significantly different than SSC (Tukey's HSD $p < 0.05$)

REFERENCES

1. Agel J, Arendt EA, Bershadsky B. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med.* 2005;33(4):524-530.PMID: 15722283 Provider: OCLC

2. Andrews M, Noyes FR, Hewett TE, Andriacchi TP. Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: a critical analysis of the reliability of gait analysis data. *J Orthop Res.* Mar 1996;14(2):289-295.PMID: 8648508 Provider: OCLC

3. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. *Am J Sports Med.* Nov-Dec 1995;23(6):694-701.8600737

4. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* Jun 2000;23(6):573-578.PMID: 10875418; Medline: 20331673 Provider: OCLC

5. Bowsher KA, Vaughan CL. Effect of foot-progression angle on hip joint moments during gait. *J Biomech.* Jun 1995;28(6):759-762.PMID: 7601876 Provider: OCLC

6. Daffertshofer A, Lamoth CJ, Meijer OG, Beek PJ. PCA in studying coordination and variability: a tutorial. *Clin Biomech (Bristol, Avon).* May 2004;19(4):415-428.PMID: 15109763 Provider: OCLC

7. Davis I, Ireland ML, Hanaki S. ACL injuries--the gender bias. *J Orthop Sports Phys Ther.* Feb 2007;37(2):A2-7.PMID: 17366963 Provider: OCLC

8. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech.* 1996;29(9):1223-1230.PMID: 8872282 Provider: OCLC

9. Ferber R, Davis IM, Williams DS. Gender differences in lower extremity mechanics during running. *Clin Biomech.* 2003;18:350-357.PMID: 12689785 Provider: OCLC

10. Forner-Cordero A, Levin O, Li Y, Swinnen SP. Principal component analysis of complex multijoint coordinative movements. *Biol Cybern.* Jul 2005;93(1):63-78.PMID: 16021517 Provider: OCLC

11. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng.* 1983;105(2):136-144.PMID: 6865355 Provider: OCLC

12. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, van den Bogert AJ, Paterno MV, Succop P. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* 2005;33:492-501.PMID: 15722287 Provider: OCLC

13. Kirkendall DT, Garrett WE. The anterior cruciate ligament enigma. Injury mechanisms and prevention. *Clinical orthopaedics and related research.* 2000;372:64-68.PMID: 10738415 Provider: OCLC

14. Maletius W, Messner K. Eighteen- to twenty-four-year follow-up after complete rupture of the anterior cruciate ligament. *Am J Sports Med.* 1999;27(6):711-717.PMID: 10569355 Provider: OCLC

15. Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of knee joint motion patterns between men and women in selected athletic tasks. *Clinical biomechanics.* Jun 2001;16(5):438-445.PMID: 11390052; Medline: 21285965 Provider: OCLC

16. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, J.l. S. Combined knee loading states that generate high anterior cruciate ligament forces. *Orhop Res. Society.* 1995;13:930-935.PMID: 8544031 Provider: OCLC

17. Markolf KL, Gorek JF, Kabo JM, Shapiro MS. Direct measurement of resultant forces in the anterior cruciate ligament. An in vitro study performed with a new experimental technique. *J Bone Joint Surg Am).* 1990;72:557-567.PMID: 2324143 Provider: OCLC

18. McClay Davis I, Ireland MI. ACL injuries--the gender bias. *J Orthop Sports Phys Ther.* 2003;33(8):A2-8.PMID: 12968861 Provider: OCLC

19. McLean SG, Huang X, Su A, Van Den Bogert AJ. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clin Biomech.* 2004;19(8): 828-838.PMID: 15342155 Provider: OCLC

20. McLean SG, Huang X, van den Bogert AJ. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech.* 2005;20(8):863-870.PMID: 16005555 Provider: OCLC

21. McLean SG, Lipfert SW, van den Bogert AJ. Effect of gender and defensive opponent on the biomechanics of sidestep cutting. 2004.PMID: 15179171 Provider: OCLC

22. McLean SG, Walker KB, van den Bogert AJ. Effect of gender on lower extremity kinematics during rapid direction changes: an integrated analysis of three sports movements. *J Sci Med Sport.* Dec 2005;8(4):411-422.PMID: 16602169 Provider: OCLC

23. McNair PJ, Marshall RN, Matheson JA. Important features associated with acute anterior cruciate ligament injury. *N Z Med J.* 1990;103:537-539.PMID: 2243642 Provider: OCLC

24. Pollard CD, Davis IM, Hamill J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech.* 2004;19(10):1022-1031.PMID: 15531052 Provider: OCLC

25. Pollard CD, Sigward SM, Powers CM. Gender differences in hip joint kinematics and kinetics during side-step cutting maneuver. *Clin J Sport Med.* Jan 2007;17(1):38-42.PMID: 17304004 Provider: OCLC

26. Powers CM. The Influence of Altered Lower-Extremity Kinematics on Patellofemoral Joint Dysfunction: A Theoretical Perspective. *J Orthop Sports Phys Ther.* 2003;33:693-646

27. Romani W, Patrie J, Curl LA, Flaws JA. The correlations between estradiol, estrone, estriol, progesterone, and sex hormone-binding globulin and anterior cruciate ligament stiffness in healthy, active females. *J Womens Health.* 2003;12(3):287-298.PMID: 12804359 Provider: OCLC

28. Shultz SJ, Kirk SE, Johnson ML, Sander TC, Perrin DH. Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Med Sci Sports Exerc.* 2004;36(7):1165-1174.PMID: 15235320 Provider: OCLC

29. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. *Clin Biomech.* 2006;21:41-48.PMID: 16209900 Provider: OCLC

30. Vaughan CL, Davis, B. L., O'Conner, J. C. *Dynamics of Human Gait*. 2nd ed. Cape Town, South Africa: Kiboho Publishers; 1999.

31. Wascher DC, Markolf KL, Shapiro MS, Finerman GA. Direct in vitro measurement of forces in the cruciate ligaments. Part I: The effect of multiplane loading in the intact knee. *J Bone Joint Surg Am.* 1993;75:377-386.PMID: 8444916 Provider: OCLC

32. Winter DA. *Biomechanics and motor control of human movement*. Second ed. Toronto: Wiley-Interscience; 1990.

GENERAL CONCLUSION

One of the primary findings of this study supports the premise that lower extremity biomechanics significantly differ between running and change-of-direction tasks. Females appear to differ in hip mechanics compared to males; however, knee mechanics were not shown to differ as a function of sex during running and change of direction tasks. Static lower extremity alignment and hip strength were not shown to contribute to the explanation of differences in knee loading and position across these same tasks. Additionally, lower extremity alignment and hip strength were not found to differ between the male and female athletes participating in this study. The coupling of knee and hip kinematics were different between running and change-of-direction tasks and this finding may help to explain how the 'at risk' maneuvers are associated with anterior cruciate ligament injury.

Results from this investigation are in agreement with other descriptive studies of running, supporting evidence of differences in hip adduction between females and males.[35] Additionally the data confirm knee loading in the frontal plane does not differ between females and males during running. However, the results do not support previously reported differences in knee abduction as a function of sex during running.[35]

The data show that a change-of-direction maneuver in which the task is initiated by stepping laterally away from the intended direction of travel results in greater frontal plane abduction angles and adduction moments at the kneee when compared to the more commonly investigated sidestep cut. While knee abduction has been implicated as one of the primary predictors of ACL injury during jump landing tasks[54], there is still a lack of consensus regarding how this factor is related to injury during change-of-direction tasks. Previous investigations of sidestep cutting have led to the assertion that the relative knee abduction associated with sidestep cutting may not be great enough to directly elicit a non-contact ACL injury.[91] The combination of a lateral false step with a sidestep cut represents one way to significantly increase knee abduction. This result may possibly explain how athletes are at risk of injury when changing direction.

The results of this study affirm other reports of a lack of a difference in frontal plane knee kinematics between females and males during rapid change-of-direction tasks.[107, 118] However, females and males were not found to differ in knee loading patterns across tasks, which deviates from other investigations of similar tasks.[91, 119] These dissimilarities suggest further investigation is warranted in order to further determine the extent of any differences in frontal plane knee loading between females and males. Females were found to demonstrate significantly different hip mechanics when compared to males. In particular, females had greater overall hip adduction and smaller external rotation moments compared to males. Although the results of this study do not make it possible to attribute the source of these differences to a specific factor, it is plausible females and males differ in strength or activation patterns of selected hip musculature. These results imply that control of the joint proximal to the knee may be related to knee injury risk and may help to explain why injury incidence differs as a function of sex.

The hip abductors contribute to the control of hip adduction and rotation.[31] Intuitively, a lack of strength of the hip abductors would result in differences in hip and knee mechanics due to the contribution of these muscles to femur position. The results of this study did not support evidence of an association between eccentric hip abductor strength and frontal plane knee position and loading during running and rapid change-of-direction tasks. Therefore, either the method used to measure hip strength was not appropriate for determining the existence of this relationship, or other hip musculature may be better associated with knee biomechanics during these tasks. For example, hip extensors, such as the gluteus maximus which extends and rotates the hip, may be more important in explaining lower extremity loading. Muscle activation patterns were not measured in this study and it is possible females differ from males with respect to when hip muscles are engaged during dynamic tasks. Researchers should continue to consider what other factors contribute to the control of hip position in order to help explain different lower extremity biomechanics associated with different tasks and between females and males.

The suggestion of static lower extremity alignment being a predictor of frontal plane knee biomechanics is appealing. The ease of measurement of standing Q-angle makes it a practical tool for clinicians to screen athletes with respect to risk of knee injury. Although athletes with abnormally large Q-angle may be at risk of overuse soft tissue injuries of the lower extremity, as it is believed to contribute to knee extensor dysfunction[77], the results from this study indicate it is not as valuable in predicting lower extremity alignment during change-of-direction tasks. This finding is in agreement with other investigations that have evaluated the relationship between standing Q-angle and knee biomechanics during a single-leg squat.[105] While it has been shown standing Q-angle is a predictor of ACL injury[100], our data suggest this relationship may not be due to the aefect of this measurement on knee position. Therefore, it is necessary to determine which risk factor this static measurement is related to in order to explain how it contributes to ACL injury risk. Despite overwhelming evidence of differences in standing Q-angle between females and males in the general population,[48, 59, 131] the results of this study suggest female and males matched for sport and age demonstrate similarities in this measurement. This may in part help explain why there is a lack of evidence of a relationship between standing Q-angle and frontal plane knee position and loading in this study, since there was not a differences in knee abduction and internal knee adduction moments as a function of sex across the investigated tasks.

Unique to this study was the description of patterns of knee and hip mechanics during running and change-of-direction tasks. Previously, researchers have measured peak kinematics and kinetics of the knee and hip across stance during running and sidestep cutting and direct relationships between these variables are assumed. It stands to reason that knee and hip position and loading behaviors are highly inter-related but the precise relationships have not been quantified. A large percentage of the variance of lower extremity kinematic and kinetic behaviors during running and change-of-direction tasks was explained by four distinct patterns of biomechanical coupling.

Normal running is not likely to result in ACL injury. The results from this study help to explain this based upon which lower extremity biomechanical factors are believed to be associated with the ACL loading. Running is associated with greater knee extension than change-of-direction tasks. The results of this study provide evidence of a pattern in which knee extension is coupled with less knee abduction and loading of the knee and hip in the sagittal, frontal and transverse planes in such a manner as to protect the ACL.

Rapid change-of-direction tasks, such as the SSC and LFS-SSC, demonstrated greater presence of a pattern in which hip abduction was related to knee adduction loading and hip adduction and external rotation loading. Specifically, in the presence of greater hip abduction, knee adduction moments were also greater. Researchers have proposed static structures are more responsible for providing support during abduction-adduction loading of the knee than musculature.[78] Therefore, we may conclude the ACL and other non-contractile structures play an important role in contributing to the knee adduction moment observed during these dynamic tasks.

The third pattern represented in the data associated hip flexion with knee flexion and hip extension moments. The relationships between these factors indicate that, in the presence of smaller hip flexion angles, the protection offered by knee flexion and hip extension loading is reduced. Although the presence of this pattern was not found to differ across tasks, it may help to explain one means by which the ACL is more susceptible to injury. Athletes, who perform rapid change-of-direction tasks in a more upright posture relative to the hip may be at greater risk of ACL injury. While it is not know what parameters of an athlete's biomechanical behavior will tend to decrease hip flexion during these tasks, clinicians may use this information to guide evaluation and intervention when observing the performance of these tasks.

The last pattern identified during this study indicated that the presence of foot abduction (externally rotated foot progression angle), was related to smaller knee and hip rotation positions as well as smaller hip rotation moments. Increases in foot abduction and external hip rotation were coupled with decreases in the angle and moment for knee internal rotation. This may suggest a relatively externally rotated lower extremity may be protective of the ACL. Although other researchers have suggested external hip rotation is related to knee abduction, our results did not provide evidence of such a relationship.

In conclusion, the combination of a lateral false step with sidestep cutting has a greater negative affect on knee and hip biomechanics than does sidestep cutting alone, in such a manner as to potentially increase loading on the ACL and secondarily place the ACL at risk of injury. While differences between females and males in knee biomechanics across these tasks were not evident, it appears hip biomechanics are associated with a gender bias. Females may be at greater risk of ACL injury due to differences in hip control when compared to males. The proposed relationships between relative hip strength and lower extremity alignment with knee position and loading were not supported in this study. The extent to which coupled knee and hip kinematic and kinetic measures are present appears to differ according to the type of dynamic task performed in such a way as to help explain why rapid change-of-direction tasks are more likely to results in ACL injury than running. These results indicate variations of the sidestep cutting task should be further evaluated and considered as possible mechanisms for ACL injury. Other factors associated with the control of hip position should be considered for further analysis in order to help explain why females are at greater risk of ACL injury than males.

BIBLIOGRAPHY

1. Agel J, Arendt EA, Bershadsky B. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med.* 2005;33(4):524-530.PMID: 15722283 Provider: OCLC

2. Anderson AF, Dome DC, Gautam S, Awh MH, Rennirt GW. Correlation of anthropometric measurements, strength, anterior cruciate ligament size, and intercondylar notch characteristics to sex differences in anterior cruciate ligament tear rates. *Am J Sports Med.* Jan-Feb 2001;29(1):58-66.11206258

3. Andrews JR, McLeod WD, Ward T, Howard K. The cutting mechanism. *Am J Sports Med.* 1977;5(3):111-121.PMID: 860773 Provider: OCLC

4. Andrews M, Noyes FR, Hewett TE, Andriacchi TP. Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: a critical analysis of the reliability of gait analysis data. *J Orthop Res.* Mar 1996;14(2):289-295.PMID: 8648508 Provider: OCLC

5. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. *Am J Sports Med.* Nov-Dec 1995;23(6):694-701.8600737

6. Arendt EA, Bershadsky B, Agel J. Periodicity of noncontact anterior cruciate ligament injuries during the menstrual cycle. *J Gend Specif Med.* 2002;5(2):19-26.PMID: 11974671 Provider: OCLC

7. Belanger MJ, Moore DC, Crisco JJ, 3rd, Fadale PD, Hulstyn MJ, Ehrlich MG. Knee laxity does not vary with the menstrual cycle, before or after exercise. *Am J Sports Med.* 2004;32(5):1150-1157.PMID: 15262636 Provider: OCLC

8. Besier TF, Lloyd DG, Ackland TR. Muscle activation strategies at the knee during running and cutting maneuvers. *Med Sci Sports Exerc.* 2003;35(1):119-127.PMID: 12544645 Provider: OCLC

9. Besier TF, Lloyd DG, Ackland TR, Cochrane JL. Anticipatory effects on knee joint loading during running and cutting maneuvers. *Med Sci Sports Exerc.* 2001;33(7):1176-1181.PMID: 11445765 Provider: OCLC

10. Besier TF, Lloyd DG, Cochrane JL, Ackland TR. External loading of the knee joint during running and cutting maneuvers. *Med Sci Sports Exerc.* 2001;33(7):1168-1175.PMID: 11445764 Provider: OCLC

11. Beynnon BD, Bernstein IM, Belisle A, Brattbakk B, Devanny P, Risinger R, Durant D. The effect of estradiol and progesterone on knee and ankle joint laxity. *Am J Sports Med.* 2005;33(9):1298-1304.PMID: 16002485 Provider: OCLC

12. Beynnon BD, Fleming BC. Anterior cruciate ligament strain in-vivo: a review of previous work. *J Biomech.* 1998;31(6):519-525.PMID: 9755036 Provider: OCLC

13. Beynnon BD, Fleming BC, Johnson RJ, Nichols CE, Renström PA, Pope MH. Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo. *Am J Sports Med.* 1995;23(1):24-34.PMID: 7726347 Provider: OCLC

14. Biedert RM, Stauffer E, Friederich NF. Occurrence of free nerve endings in the soft tissue of the knee joint. A histologic investigation. *Am J Sports Med.* 1992;20(4):430-433.PMID: 1415886 Provider: OCLC

15. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* Jun 2000;23(6):573-578.PMID: 10875418; Medline: 20331673 Provider: OCLC

16. Bowsher KA, Vaughan CL. Effect of foot-progression angle on hip joint moments during gait. *J Biomech.* Jun 1995;28(6):759-762.PMID: 7601876 Provider: OCLC

17. Butler DL, Noyes FR, Grood ES. Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. *J Bone Joint Surg Am.* 1980;62(2):259-270.PMID: 7358757 Provider: OCLC

18. Byl T, Cole JA, Livingston LA. What determines the magnitude of the Q angle? A preliminary study of selected skeletal and muscular measures. *J Sport Rehab.* 2000;9:26-34

19. Cahalan TD, Johnson ME, Liu S, Chao EY. Quantitative measurements of hip strength in different age groups. *Clin Orthop Relat Res.* 1989;246:136-145.PMID: 2766602 Provider: OCLC

20. Chandrashekar N, Mansouri H, Slauterbeck J, Hashemi J. Sex-based differences in the tensile properties of the human anterior cruciate ligament. *American Journal of Sports Medicine.* 2006;39:2943-2950.PMID: 16387307 Provider: OCLC

21. Chandrashekar N, Slauterbeck J, Hashemi J. Sex-based differences in the anthropometric characteristics of the anterior cruciate ligament and its relation to intercondylar notch geometry: a cadaveric study. *American Journal of Sports Medicine.* 2005;33:1492-1498.PMID: 16009992 Provider: OCLC

22. Chaudhari AM, Andriacchi TP. The mechanical consequences of dynamic frontal plane limb alignment for non-contact ACL injury. *J Biomech.* 2006;39(2):330-338.PMID: 16321635 Provider: OCLC

23. Claiborne TL, Armstrong CW, Gandhi V, Pincivero DM. Relationship between hip and knee strength and knee valgus during a single leg squat. *J Appl Biomech.* Feb 2006;22(1):41-50.PMID: 16760566 Provider: OCLC

24. Colby S, Francisco A, Yu B, Kirkendall D, Finch M, W. G. Electromyographic and kinematic analysis of cutting maneuvers. Implications for anterior cruciate ligament injury. *Am J Sports Med.* 2000;28(2):234-240.PMID: 10751001 Provider: OCLC

25. Cross MJ, Gibbs NJ, Bryant GJ. An analysis of the sidestep cutting manoeuvre. *The American journal of sports medicine.* May-Jun 1989;17(3):363-366.PMID: 2729486; Medline: 89270834 Provider: OCLC

26. Cross MJ, Gibbs NJ, Bryant GJANSO, Sports Medicine Center CNNSWA. An analysis of the sidestep cutting manoeuvre. 1989.PMID: 2729486 Provider: OCLC

27. Daffertshofer A, Lamoth CJ, Meijer OG, Beek PJ. PCA in studying coordination and variability: a tutorial. *Clin Biomech (Bristol, Avon).* May 2004;19(4):415-428.PMID: 15109763 Provider: OCLC

28. Davis I, Ireland ML, Hanaki S. ACL injuries--the gender bias. *J Orthop Sports Phys Ther.* Feb 2007;37(2):A2-7.PMID: 17366963 Provider: OCLC

29. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech.* 1996;29(9):1223-1230.PMID: 8872282 Provider: OCLC

30. Deie M, Sakamaki Y, Sumen Y, Urabe Y, Ikuta Y. Anterior knee laxity in young women varies with their menstrual cycle. *Int Orthop.* 2002;26(3):154-156.PMID: 12073107 Provider: OCLC

31. Delp SL, Hess WE, Hungerford DS, Jones LC. Variation of rotation moment arms with hip flexion. *J Biomech.* 1999;32(5):493-501.PMID: 10327003 Provider: OCLC

32. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med.* Mar 2004;32(2):477-483.PMID: 14977677 Provider: OCLC

33. Dienst M, Burks RT, Greis PE. Anatomy and biomechanics of the anterior cruciate ligament. *Orthop Clin North Am.* 2002;33(4):605-620.PMID: 12528904 Provider: OCLC

34. Fagenbaum R, Darling WG. Jump landing strategies in male and female college athletes and the implications of such strategies for anterior cruciate ligament injury. *Am J Sports Med.* Mar-Apr 2003;31(2):233-240.12642258

35. Ferber R, Davis IM, Williams DS. Gender differences in lower extremity mechanics during running. *Clin Biomech.* 2003;18:350-357.PMID: 12689785 Provider: OCLC

36. Fischer-Rasmussen T, Jensen PE. Proprioceptive sensitivity and performance in anterior cruciate ligament-deficient knee joints. *Scand J Med Sci Sports.* 2000;10(2):85-89.PMID: 10755278 Provider: OCLC

37. Ford KR, Myer GD, Toms HE, Hewett TE. Gender differences in the kinematics of unanticipated cutting in young athletes. *Med Sci Sports Exerc.* 2005;37:124-129.PMID: 15632678 Provider: OCLC

38. Forner-Cordero A, Levin O, Li Y, Swinnen SP. Principal component analysis of complex multijoint coordinative movements. *Biol Cybern.* Jul 2005;93(1):63-78.PMID: 16021517 Provider: OCLC

39. Frank CB, Jackson DW. The science of reconstruction of the anterior cruciate ligament. *J Bone Joint Surg Am.* 1997;79(10):1556-1576.PMID: 9378743 Provider: OCLC

40. Fredericson M, Cookingham CL, Chaudhari AM, Dowdell BC, Oestreicher N, Sahrmann SA. Hip abductor weakness in distance runners with iliotibial band syndrome. *Clin J Sport Med.* Jul 2000;10(3):169-175.PMID: 10959926 Provider: OCLC

41. Freeman MA, Wyke B. Articular contributions to limb muscle reflexes. The effects of partial neurectomy of the knee-joint on postural reflexes. *Br J Surg.* 1966;53(1):61-68.PMID: 5901916 Provider: OCLC

42. Fung DT, L.Q. Z. Modeling of ACL impingement against the intercondylar notch. *Clin Biomech.* 2003;18(10):933-941.PMID: 14580837 Provider: OCLC

43. Golden GM. *The effect of altering the width of a single step of a running stride on knee joint kinematics and kinetics* [In Lieu of Thesis, unpublished manuscript]. Corvallis: Nutrition and Exercise Sciences, Oregon State University; 2005.

44. Golden GM, Pavol MJ, Hoffman MA. *Sex differences in kinematics and kinetics of the knee and hip during running and rapid change of direction tasks* [Dissertation]. Corvallis: Nutrition and Exercise Sciences, Oregon State University; 2007.

45. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, Garrick JG, Hewett TE, Huston L, Ireland ML, Johnson RJ, Kibler WB, Lephart S, Lewis JL, Lindenfeld TN, Mandelbaum BR, Marchak P, Teitz CC, Wojtys EM. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *The Journal of the American Academy of Orthopaedic Surgeons.* May-Jun 2000;8(3):141-150.PMID: 10874221; Medline: 20334910 Provider: OCLC

46. Griffin LY, Albohm MJ, Arendt EA, Bahr R, Beynnon BD, Demaio M, Dick RW, Engebretsen L, Garrett WE, Jr., Hannafin JA, Hewett TE, Huston LJ, Ireland ML, Johnson RJ, Lephart S, Mandelbaum BR, Mann BJ, Marks PH, Marshall SW, Myklebust G, Noyes FR, Powers C, Shields C, Jr., Shultz SJ, Silvers H, Slauterbeck J, Taylor DC, Teitz CC, Wojtys EM, Yu B. Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *American Journal of Sports Medicine.* 2006;34:1512-1532.PMID: 16905673 Provider: OCLC

47. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng.* 1983;105(2):136-144.PMID: 6865355 Provider: OCLC

48. Guerra JP, Arnold MJ, Gajdosik RL. Q angle: effects of isometric quadriceps contraction and body position. *J Orthop Sports Phys Ther.* Apr 1994;19(4):200-204.PMID: 8173567 Provider: OCLC

49. Harmon KG, Ireland ML. Gender differences in noncontact anterior cruciate ligament injuries. *Clinics in sports medicine.* Apr 2000;19(2):287-302.PMID: 10740760; Medline: 20204933 Provider: OCLC

50. Heitz NA, Eisenman PA, Beck CL, Walker JA. Hormonal changes throughout the menstrual cycle and increased anterior cruciate ligament laxity in females. *J Athl Train.* 1999;34(2):144-149

51. Hewett TE. Neuromuscular and hormonal factors associated with knee injuries in female athletes. Strategies for intervention. *Sports Med.* 2000;29(5):313-327.PMID: 10840866; Medline: 20299904 Provider: OCLC

52. Hewett TE, Myer GD, Ford KR. Prevention of anterior cruciate ligament injuries. *Curr Womens Health Rep.* 2001;1(3):218-224.PMID: 12112973 Provider: OCLC

53. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, van den Bogert AJ, Paterno MV, Succop P. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* 2005 2005;33(4):492-501.PMID: 15722287 Provider: OCLC

54. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, van den Bogert AJ, Paterno MV, Succop P. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* 2005;33:492-501.PMID: 15722287 Provider: OCLC

55. Hewett TE, Stroupe AL, Nance TA, Noyes FRACS, Orthopaedic Center OUSA. Plyometric training in female athletes. Decreased impact forces and increased hamstring torques. 1996.PMID: 8947398; Medline: 97103026 Provider: OCLC

56. Hirokawa S, Solomonow M, Lu Y, Lou ZP, D'Ambrosia R. Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction. *Am J Sports Med.* 1992;20(3): 299-306.PMID: 1636861 Provider: OCLC

57. Hollman JH, Kolbeck KE, Hitchcock JL, Koverman JW, Krause DA. Correlations between hip strength and static foot and knee posture. *J Sport Rehab.* 2006;15:12-23

58. Horton MG, Hall TL. Quadriceps femoris muscle angle: normal values and relationships with gender and selected skeletal measures. *Phys Ther.* 1989;69(11):897-901.PMID:

59. Horton MG, Hall Tl. Quadriceps femoris muscle angle: normal values and relationships with gender and selected skeletal measures. *Phys Ther.* 1989;69:897-901.PMID: 2813517 Provider: OCLC

60. Huston LJ, Greenfield ML, Wojtys EM. Anterior cruciate ligament injuries in the female athlete. Potential risk factors. *Clin Orthop Relat Res.* 2000;372:50-63.PMID: 10738414 Provider: OCLC

61. Huston LJ, Wojtys EM. Neuromuscular performance characteristics in elite female athletes. *Am J Sports Med.* 1996;24(4):427-436.PMID: 8827300 Provider: OCLC

62. Hutchinson MR, Ireland ML. Knee injuries in female athletes. *Sports medicine.* Apr 1995;19(4):288-302.PMID: 7604201; Medline: 95327860 Provider: OCLC

63. Ireland ML. The female ACL: why is it more prone to injury? *The Orthopedic clinics of North America.* Oct 2002;33(4):637-651.PMID: 12528906; Medline: 22416240 Provider: OCLC

64. Jacobs C, Mattacola C. Sex differences in eccentric hip-abductor strength and knee-joint kinematics when landing from a jump. *J Sport Rehabil.* 2005;14:346-355

65. Jacobs CA, Uhl TL, Mattacola CG, Shapiro R, Rayens WS. Hip abductor function and lower extremity landing kinemtaics: sex differences. *Journal of Athletic Training.* 2007;42:76-83

66. Johansson H, Sjölander P, Sojka P. Receptors in the knee joint ligaments and their role in the biomechanics of the joint. *Crit Rev Biomed Eng.* 1991;18(5):341-368.PMID: 2036801 Provider: OCLC

67. Kanamori A, Sakane M, Zeminski J, Rudy TW, S.l. W. In-situ force in the medial and lateral structures of intact and ACL-deficient knees. *J Orthop Sci.* 2000;5(6):567-571.PMID: 11180920 Provider: OCLC

68. Karageanes SJ, Blackburn K, Vangelos ZA. The association of the menstrual cycle with the laxity of the anterior cruciate ligament in adolescent female athletes. *Clin J Sport Med.* 2000;10(3):162-168.PMID: 10959925 Provider: OCLC

69. Kirkendall DT, Garrett WE. The anterior cruciate ligament enigma. Injury mechanisms and prevention. *Clinical orthopaedics and related research.* 2000;372:64-68.PMID: 10738415 Provider: OCLC

70. Lathinghouse LH, Trimble MH. Effects of isometric quadriceps activation on the Q-angle in women before and after quadriceps exercise. *J Orthop Sports Phys Ther.* Apr 2000;30(4):211-216.PMID: 10778798 Provider: OCLC

71. Lebrun CM. Effect of the different phases of the menstrual cycle and oral contraceptives on athletic performance. *Sports Med.* 1993;16(6):400-430.PMID: 8303141 Provider: OCLC

72. Lebrun CM. The effect of the phase of the menstrual cycle and the birth control pill on athletic performance. *Clin Sports Med.* 1994;13(2):419-441.PMID: 8013042 Provider: OCLC

73. Leetun DT, Ireland ML, Willson JD, Ballantyne BT, Davis IM. Core stability measures as risk factors for lower extremity injury in athletes. *Med Sci Sports Exerc.* Jun 2004;36(6):926-934.PMID: 15179160 Provider: OCLC

74. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo Sl. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech.* 1999;32(4):395-400.PMID: 10213029 Provider: OCLC

75. Liu SH, al-Shaikh R, Panossian V, Yang RS, Nelson SD, Soleiman N, Finerman GA, Lane JM. Primary immunolocalization of estrogen and progesterone target cells in the human anterior cruciate ligament. *J Orthop Res.* 1996;14(4):526-533.PMID:

76. Liu SH, Al-Shaikh RA, Panossian V, Finerman GA, Lane JM. Estrogen affects the cellular metabolism of the anterior cruciate ligament. A potential explanation for female athletic injury. *Am J Sports Med.* 1997;25(5):704-709.PMID: 9302481 Provider: OCLC

77. Livingston LA. The quadriceps angle: a review of the literature. *J Orthop Sports Phys Ther.* 1998;28(2):105-109.PMID: 9699160 Provider: OCLC

78. Lloyd DG, Buchanan TS. A model of load sharing between muscles and soft tissues at the human knee during static tasks. *J Biomech Eng.* Aug 1996;118(3):367-376.PMID: 8872259 Provider: OCLC

79. Lloyd DG, Buchanan TS. Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech.* 2001;34(10):1257-1267.PMID: 11522305 Provider: OCLC

80. Lovering RM, Romani WA. Effect of testosterone on the female anterior cruciate ligament. *Am J Physiol Regul Integr Comp Physio.* 2005;289(1):R15-22.PMID: 15790748 Provider: OCLC
81. Maletius W, Messner K. Eighteen- to twenty-four-year follow-up after complete rupture of the anterior cruciate ligament. *Am J Sports Med.* 1999;27(6):711-717.PMID: 10569355 Provider: OCLC
82. Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of knee joint motion patterns between men and women in selected athletic tasks. *Clinical biomechanics.* Jun 2001;16(5):438-445.PMID: 11390052; Medline: 21285965 Provider: OCLC
83. Mansour JM, Pereira JM. Quantitative functional anatomy of the lower limb with application to human gait. *J Biomech.* 1987;20(1):51-58.PMID: 3558428 Provider: OCLC
84. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, J.l. S. Combined knee loading states that generate high anterior cruciate ligament forces. *Orhop Res. Society.* 1995;13:930-935.PMID: 8544031 Provider: OCLC
85. Markolf KL, Gorek JF, Kabo JM, Shapiro MS. Direct measurement of resultant forces in the anterior cruciate ligament. An in vitro study performed with a new experimental technique. *J Bone Joint Surg Am).* 1990;72:557-567.PMID: 2324143 Provider: OCLC
86. McClay Davis I, Ireland MI. ACL injuries--the gender bias. *J Orthop Sports Phys Ther.* 2003;33(8):A2-8.PMID: 12968861 Provider: OCLC
87. McClay Davis I, Ireland ML. ACL research retreat: the gender bias. April 6-7, 2001. Meeting report and abstracts. *Clinical biomechanics (Bristol, Avon).* Dec 2001;16(10):937-959.PMID: 11733134; Medline: 21590278 Provider: OCLC
88. McClay IS, Robinson JR, Andriacchi TP, Frederick TG, Martin P, Valiant G, Williams KR, Cavanagh PR. A kinematic profile of skills in professional basketball players. *J Appl Biomech.* 1994;10:205-221
89. McLean SG, Andrish JT, van den Bogert AJ, Garrett WE, Jr., Yu B. Letter to the editor * authors' response. 2005.PMID: 15983128 Provider: OCLC
90. McLean SG, Huang X, Su A, Van Den Bogert AJ. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clin Biomech.* 2004;19(8): 828-838.PMID: 15342155 Provider: OCLC
91. McLean SG, Huang X, van den Bogert AJ. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech.* 2005;20(8):863-870.PMID: 16005555 Provider: OCLC
92. McLean SG, Lipfert SW, van den Bogert AJ. Effect of gender and defensive opponent on the biomechanics of sidestep cutting. 2004.PMID: 15179171 Provider: OCLC
93. McLean SG, Myers PT, Neal RJ, Walters MR. A quantitative analysis of knee joint kinematics during the sidestep cutting maneuver. Implications for non-contact anterior cruciate ligament injury. *Bull Hosp Jt Dis.* 1998;57(1):30-38.PMID: 9553700 Provider: OCLC
94. McLean SG, Neal RJ, Myers PT, Walters MR. Knee joint kinematics during the sidestep cutting maneuver: potential for injury in women. *Med Sci Sports Exerc.* 1999;31(7):959-968.PMID: 10416556 Provider: OCLC
95. McLean SG, Walker KB, van den Bogert AJ. Effect of gender on lower extremity kinematics during rapid direction changes: an integrated analysis of three sports movements. *J Sci Med Sport.* Dec 2005;8(4):411-422.PMID: 16602169 Provider: OCLC
96. McNair PJ, Marshall RN, Matheson JA. Important features associated with acute anterior cruciate ligament injury. *The New Zealand medical journal.* Nov 14 1990;103(901):537-539.PMID: 2243642; Medline: 91056954 Provider: OCLC
97. McNair PJ, Marshall RN, Matheson JA. Important features associated with acute anterior cruciate ligament injury. *N Z Med J.* 1990;103:537-539.PMID: 2243642 Provider: OCLC
98. Messier SP, Davis SE, Curl WW, Lowery RB, Pack RJ. Etiologic factors associated with patellofemoral pain in runners. *Med Sci Sports Exerc.* Sep 1991;23(9):1008-1015.PMID: 1943620 Provider: OCLC
99. Miyasaka KC, Daniel DM, Stone ML, Hirsham P. The incidence of knee ligament injuries in the general population. *Am J Knee Surg.* 1991;4(1):3-8
100. Moul J. Differences in selected predictors of anterior cruciate ligament tears between male and female NCAA Division I collegiate basketball players. *Journal of Athletic Training.* 1998/04//1998 Apr-Jun 1998;33(2):118-121

101. Myklebust G, Maehlum S, Holm I, Bahr R. A prospective cohort study of anterior cruciate ligament injuries in elite Norwegian team handball. *Scand J Med Sci Sports.* 1998;8(3):149-153.PMID: 9659675 Provider: OCLC

102. Norkin CC, Levangie PM. *Joint Structure and Function: A comprehensive analysis*. 2nd ed. Philadelphia: F.A. Davis Company; 1992.

103. Olsen OE, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med.* 2004;32(4):1002-1012.PMID: 15150050 Provider: OCLC

104. Pandy MG, Shelburne KB. Dependence of cruciate-ligament loading on muscle forces and external load. *J Biomech.* 1997;30(10):1015-1024.PMID: 9391868 Provider: OCLC

105. Pantano KJ, White SC, Gilchrist LA, Leddy J. Differences in peak knee valgus angles between individuals with high and low Q-angles during a single limb squat. *Clin Biomech (Bristol, Avon).* Nov 2005;20(9):966-972.PMID: 16051403 Provider: OCLC

106. Pap G, Machner A, Nebelung W, Awiszus F. Detailed analysis of proprioception in normal and ACL-deficient knees. *J Bone Joint Surg Br.* 1999;81(5):764-768.PMID: 10530833 Provider: OCLC

107. Pollard CD, Davis IM, Hamill J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech.* 2004;19(10):1022-1031.PMID: 15531052 Provider: OCLC

108. Pollard CD, Sigward SM, Powers CM. Gender differences in hip joint kinematics and kinetics during side-step cutting maneuver. *Clin J Sport Med.* Jan 2007;17(1):38-42.PMID: 17304004 Provider: OCLC

109. Posthuma BW, Bass MJ, Bull SB, Nisker JA. Detecting changes in functional ability in women with premenstrual syndrome. *Am J Obstet Gynecol.* 1987;156(2):275-278.PMID: 3826160 Provider: OCLC

110. Powers CM. The Influence of Altered Lower-Extremity Kinematics on Patellofemoral Joint Dysfunction: A Theoretical Perspective. *J Orthop Sports Phys Ther.* 2003;33:693-646

111. Romani W, Patrie J, Curl LA, Flaws JA. The correlations between estradiol, estrone, estriol, progesterone, and sex hormone-binding globulin and anterior cruciate ligament stiffness in healthy, active females. *J Womens Health.* 2003;12(3):287-298.PMID: 12804359 Provider: OCLC

112. Rozzi SL, Lephart SM, Gear WS, Fu FH. Knee joint laxity and neuromuscular characteristics of male and female soccer and basketball players. *Am J Sports Med.* May-Jun 1999;27(3):312-319.10352766

113. Rudolph KS, Fitzgerald K, Snyder-Mackler L. Restoration of dynamic stability in the ACL-deficient knee. In: Lephart S, Fu FH, eds. *Proprioception and Neuromuscular Control in Joint Stability*. USA: Human Kinetics; 2000:393-404

114. Sarwar R, Niclos BB, Rutherford OM. Changes in muscle strength, relaxation rate and fatiguability during the human menstrual cycle. *J Physiol.* 1996;493(Pt 1):267-272.PMID: 8735711 Provider: OCLC

115. Shambaugh JP, Klein A, Herbert JH. Structural measures as predictors of injury basketball players. *Med Sci Sports Exerc.* May 1991;23(5):522-527.1906569

116. Shultz SJ, Kirk SE, Johnson ML, Sander TC, Perrin DH. Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Med Sci Sports Exerc.* 2004;36(7):1165-1174.PMID: 15235320 Provider: OCLC

117. Shumway-Cook A, Woollacott M. Normal Posture Control. *Motor Control: Theory and Practical Applications*. 2nd ed. Baltimore: Lippincott Williams and Wilkins; 2001:163-191

118. Sigward S, Powers CM. The influence of experience on knee mechanics during side-step cutting in females. *Clin Biomech.* 2006;21:740-747.PMID: 16675083 Provider: OCLC

119. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. *Clin Biomech.* 2006;21:41-48.PMID: 16209900 Provider: OCLC

120. Slauterbeck JR, Fuzie SF, Smith MP, Clark RJ, Xu KT, Starch DW, Hardy DM. The menstrual cycle, sex hormones, and anterior cruciate ligament injury. *J Athl Train.* 2002;37(3):275-280

121. Solomonow M, Baratta R, Zhou BH, Shoji H, Bose W, Beck C, D'Ambrosia R. The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *Am J Sports Med.* 1987;15(3):207-213.PMID: 3618871 Provider: OCLC

122. Tsuda E, Okamura Y, Otsuka H, Komatsu T, Tokuya S. Direct evidence of the anterior cruciate ligament-hamstring reflex arc in humans. *Am J Sports Med.* 2001;29(1):83-87.PMID: 11206261 Provider: OCLC

123. Van Lunen BL, Roberts J, Branch JD, Dowling EA. Association of menstrual-cycle hormone changes with anterior cruciate ligament laxity measurements. *J Athl Train.* 2003;38(4):298-303

124. Vaughan CL, Davis, B. L., O'Conner, J. C. *Dynamics of Human Gait*. 2nd ed. Cape Town, South Africa: Kiboho Publishers; 1999.

125. Wascher DC, Markolf KL, Shapiro MS, Finerman GA. Direct in vitro measurement of forces in the cruciate ligaments. Part I: The effect of multiplane loading in the intact knee. *J Bone Joint Surg Am.* 1993;75:377-386.PMID: 8444916 Provider: OCLC

126. Willson JD, Ireland ML, Davis I. Core strength and lower extremity alignment during single leg squats. *Med Sci Sports Exerc.* May 2006;38(5):945-952.PMID: 16672849 Provider: OCLC

127. Winter DA. *Biomechanics and motor control of human movement*. Second ed. Toronto: Wiley-Interscience; 1990.

128. Wojtys EM, Huston LJ, Lindenfeld TN, Hewett TE, Greenfield Ml. Association between the menstrual cycle and anterior cruciate ligament injuries in female athletes. *Am J Sports Med.* 1998;26(5):614-619.PMID: 9784805 Provider: OCLC

129. Woo SL, Debski RE, Zeminski J, Abramowitch SD, Saw SS, J.A. F. Injury and repair of ligaments and tendons. *Annu Rev Biomed Eng.* 2000;2:83-118.PMID: 11701508 Provider: OCLC

130. Woo SL, Hollis JM, Adams DJ, Lyon RM, Takai S. Tensile properties of the human femur-anterior cruciate ligament-tibia complex. The effects of specimen age and orientation. *Am J Sports Med.* 1991;19(3):217-225.PMID: 1867330 Provider: OCLC

131. Woodland LH, Francis RS. Parameters and comparisons of the quadriceps angle of college-aged men and women in the supine and standing positions. *Am J Sports Med.* Mar-Apr 1992;20(2):208-211.PMID: 1558251 Provider: OCLC

132. Yu WD, Liu SH, Hatch JD, Panossian V, Finerman GA. Effect of estrogen on cellular metabolism of the human anterior cruciate ligament. *Clin Orthop Relat Res.* Sep 1999(366):229-238.PMID:

133. Yu WD, Panossian V, Hatch JD, Liu SH, Finerman GA. Combined effects of estrogen and progesterone on the anterior cruciate ligament. *Clin Orthop Relat Res.* Feb 2001(383):268-281.PMID: 11210964 Provider: OCLC

APPENDICES

APPENDIX A

REVIEW OF THE LITERATURE

Anterior cruciate ligament (ACL) injury has a significant impact on athletic performance. Although athletes participating in sports demanding agility performance are at risk of acute collateral ligament, cruciate ligament, and meniscal knee injuries, the consequences of ACL injury are most deleterious. In the general population there is an estimated one in 3000 individuals who sustain an anterior cruciate ligament injury per year in the United States.[60] Nearly 50,000 ACL reconstructions are performed annually, creating an approximate financial impact of 850 million dollars.[39] Pathologic knee motion due to ACL rupture, occurs 50% more often then isolated MCL injuries, supporting the belief that the ACL is the most frequently ruptured ligament in the knee.[99] At the collegiate level, documented ACL injury accounts for approximately 32% of knee ligament injuries, second only to collateral ligament injuries for the sports of basketball and soccer.[5] Although information is not readily available with regards to the number of ACL injuries at the collegiate level requiring surgical reconstruction in order return to a full pre-injury functional level of participation, athletes predominantly opt for surgical reconstruction sinces the outcomes are poorer for competition at a high level with ACL deficiency. Slight-to-moderate changes in the articulating surfaces of the knee, equivalent to osteoarthrosis occur in over 80% of individuals with ACL injury.[81] Therefore, the need for surgical reconstruction to restore knee stability, associated surgical and rehabilitation costs, significant time loss from participation, and the long term consequence of developing osteoarthritis following ACL injury further substantiate the need to determine what factors contribute to this injury.[81, 129]

The cause of ACL injury is believed to be multifactorial.[45, 49, 62, 63] In 2001, 2003, and 2006 a group of interdisciplinary researchers gathered as part of a collaborative effort to address which factors are associated with the incidence of ACL injury and plausible reasons for gender differences [86, 87]. Non-contact injury mechanisms account for 70-78% of all ACL ruptures in the athletic population.[15, 97] Arendt and Dick[5], subsequent to a 5-year period of injury surveillance through the National Collegiate Athletic Association, reported knee injuries of male and female soccer players account for 17% of all injuries for that sport. Over the same year period, knee injuries account for 15% of all injuries for male and female basketball players. These knee injury rates remained stable at the collegiate level from 1990-2002.[1]

Female soccer and basketball players are 2-4 times more likely to suffer ACL injury than males.[1, 5] and the rate of ACL injuries for females is significantly higher than males, regardless of the sport.[1] Incidence of ACL injury rates for females is 0.33 and 0.29 per 1000 exposures for the sports of soccer and basketball respectively. In comparison, the incidence of ACL injury rates for males is 0.11 and 0.08 per 1000 exposures for soccer and basketball.[1] Agel et al.[1] report ACL injury rates remain stable for male and female collegiate basketball players across a 13 year span. However, the incidence of ACL injury for male soccer players has decreased significantly whereas for female soccer players it has remained constant. Collegiate soccer players are more likely to suffer ACL injury than basketball players. These data support existence of a significant inconsistency of ACL injury rates between males and females as well as a significant effect of sport.

In effort to address injury rate and gender differences related to ACL injury, researchers have investigated and categorized the contributing factors as anatomical, hormonal, neuromuscular, and biomechanical.[86, 87] Additionally, clearly determining the possible mechanisms of non-contact ACL injury is still necessary. Anatomical characteristics are generally non-modifiable, meaning the size of the anatomical structures and relationships between the structures are intrinsic to the individual. While researchers recognize that intrinsic structural characteristics can be identified, determining whether these differences are related to the differences in ACL injury rates in particular populations is still under investigation. The possibility of hormonal influences on structural behavior and motor control offers significant potential in identifying differences between males and females. The practicality of using the knowledge of hormonal contributions to ligament disruption to prevent injuries still requires elucidation; however it may contribute to the explanation of gender differences in the rates of ACL injury. Identifying neuromuscular control strategies utilized during activity can aid in understanding how athletes produce joint position and movement. The affect of gender and maturation on neuromuscular control cannot be ignored as evidence of differences in coordinated muscle responses may have a significant impact on joint stiffness, loads, and three dimensional movement patterns of the knee. While neuromuscular control and biomechanical factors are related, the association between the two requires further exploration. Namely, does neuromuscular control drive the resulting biomechanical kinematics and kinetics or visa versa? It is more plausible that there is a delicate balance between the two and developing a clearer understanding of the relationship will help in identifying what variations in neuromuscular behavior increase the risk of ACL injury. There is greater potential to modify neuromuscular and biomechanical behavior through movement education and correction, driving the surge of investigations that consider these contributing factors and intervention programs.

Despite current contributions to the literature with regard to anatomical, hormonal, neuromuscular, and biomechanical factors associated with ACL injury, evidence of how the ACL is loaded in vivo during high risk maneuvers such as abrupt stopping, jump landings and rapid changes in direction while running is not known. Researchers are encouraged to investigate the neuromuscular and biomechanical impact of out of plane movements on the ACL. First, we must quantify the ordinary and therefore potentially optimal movement patterns for these tasks, and then identify unusual and potentially detrimental expressions of these patterns. In doing so, we may conclude that certain athletes demonstrate movement behaviors that place them at greater risk for ACL injury, regardless of gender, albeit more females may possess these characteristics in greater proportion to males.

This research study is specifically aimed at objectively measuring the effect of three different maneuvers utilized by athletes to rapidly change direction while running. As a biomechanical study, kinematic and kinetic differences between these three tasks will be identified. The three maneuvers include: 1.) lateral false step, 2.) sidestep cutting, and 3.) lateral false step performed in combination with a sidestep cutting maneuver. A lateral false step is represented by increasing the distance of a single step within a running stride in the lateral direction (as a percentage of body height) and then immediately returning to the original path of travel. In effect, it is a step into one direction without continuation of travel in the direction of the false step. This step is misleading to the opponent and serves to misdirect or evade the opponent during athletic performance.

Previous research has demonstrated that a lateral false step performed within straight ahead running produces significantly different knee positions and moments in the sagittal, frontal, and transverse planes than running alone.[43] There is a substantial amount of research devoted to sidestep cutting[8, 10, 24, 26, 82, 89, 90, 93, 94], however in these studies, the preceding movements of the individual performing the maneuver are controlled such that the subsequent change in direction is executed without hesitation or misdirection. This type of sidestep cutting maneuver is performed to rapidly accelerate the athlete past the opponent and serves as the evasive maneuver exclusively. Sidestep cutting also results in significantly different knee positions and moments compared to running in the sagittal, frontal, and transverse planes.

Out of plane positions and moments in the frontal and transverse planes are believed to contribute to ACL strain and potentially increase the risk of ACL injury. Females demonstrate greater knee frontal and transverse plane positioning of the knee than males in sidestep cutting tasks.[82, 92-94] Based upon the evidence that both lateral false step maneuvers and sidestep cutting result in potentially dangerous knee positions and knee loads, it seems appropriate to investigate the effect of a combination of both maneuvers on lower extremity position. This information could help in determining if performance of a lateral false step in combination with sidestep cutting is more hazardous then either maneuver in isolation. As athletes will commonly perform any of the three maneuvers during multi-directionally dependent sports, this information may help to better define the mechanism of injury for the ACL associated with rapid changes in direction.

Structurally, the ACL serves as primary restraint to anterior tibial translation as well as secondary restraint to tibial internal rotation and varus/valgus angulations, particularly in knee positions near full extension when an external load is applied to the knee.[17, 85] The bony origin of the ACL is 11-24 mm in diameter, attaching on the posterior part of the inner surface of the lateral femoral condyle. Traveling through the intercondylar notch in an anterior-medial and distal direction, the ACL fans as it nears insertion anterior and slightly lateral to the medial intercondylar tubercle of the tibial plateau.[33] The insertion site averages 8-12 mm in width and 14-21 mm in the anterior-posterior direction. It is generally reported the ACL is divided into two bundles, the anteriomedial (AMB) and posteriolateral (PLB) respectively, with a length of 22-41 mm. During passive knee flexion the AMB tightens and the PLB is relatively lax. As the knee approaches terminal extension the PLB tightens and the AMB remains relatively taut.[33] In vivo studies of ACL loading patterns most often consider the ACL complex as a single structure.

Objective measurement of the femoral intercondylar notch is typically determined as a ratio of epicondylar width to notch width, reported as reverse notch width index (NWI). The range of NWI has been reported to fall between 0.21 to 0.25.[33] Concerns related to NWI and ACL diameter have centered on the idea that notch width stenosis may be a predisposing factor for ACL tear. Statistically smaller NWI have been reported in patients who have suffered ACL rupture; however, evidence is less clear regarding

NWI differences between males and females as both gender related differences and a lack of gender related differences have been reported. This may be due to the fact that a large range of NWI ratios exist and there is considerable overlap between males and females.[33]

Extrinsic measurement of the tensile strength of the ACL depends upon size and shape of the structure. Typically, the structural behavior of the ACL includes measurement of the bone-ligament-bone complex and is described with a load-deformation curve. Ultimate load to failure for the femur-ACL-tibia complex is reported to be 2160 ± 157 N, with a linear stiffness of 242 ± 28 N/mm.[129] Ultimate load and linear stiffness of the complex reduces with age. In vivo studies have demonstrated that the ACL undergoes positive strain during isometric quadriceps contraction from positions of full knee extension to 40° of knee flexion, during active range of motion, clinical loads with a Lachman's test of 150N at 30° of knee flexion, and squatting.[12, 13] Although the strain response of the ACL is well documented in non-weight bearing and low level activity conditions, less information is available regarding ACL behavior under more dynamic loading. The interaction of ligament forces, joint surface contact, externally applied loads and muscle activity affords the capacity of the knee joint to withstand various types of loading and large external forces.[129] Therefore, determining how the ACL responds during locomotive activities is a more complex endeavor. Until further information is available, researchers are relegated to relating what is known about ACL strain behavior in non-weight bearing and low level weight bearing activities in order to predict how the ACL may be injured during more dynamic activity.

In addition to the ACL's role in providing mechanical stability to the knee, it has been suggested the ACL serves a proprioceptive function. This idea is largely based upon histological evidence of proprioceptive nerve ending located in the human ACL.[14, 66] In 1966 Freeman and Wyke[41] proposed that articular mechanoreceptors normally contribute significantly to postural regulation and muscle tone. Although the ACL potentially serves as a joint afferent receptor, the role of the capsule, collateral ligaments, muscle and tendon afferents cannot be ignored. The research of Solomonow et al.[121] in 1987, has led many to believe that the ACL constitutes part of a protective reflex arc, whereby mechanical stimulation of the ACL facilitates hamstring activity. More recently, Tsuda et al. [122] demonstrated electrical stimulation of the ACL results in direct increase in biceps femoris and semitendinosis hamstring muscle activity. The hamstrings are protective of the ACL, preventing anterior translation of the tibia;

therefore existence of a reflex arc is intriguing. However evidence of the significance to which the reflex arc contributes to protection of the ACL during dynamic activity is not available. Assessment of proprioception of the knee by Pap et al.[106] suggests that subjects with ACL-deficient (ACLD) knees do not have a difference in the threshold to detection of passive movement (TTDPM) compared to the contralateral limb.[106] These researchers also reported that ACLD patients have no difference in TTDPM compared to healthy controls, but do have increased incidence of failure to detect passive movement, therefore a proprioceptive deficit. Contrary these results, Fischer-Rasmussen[36], when comparing ACLD, ACL-reconstructed (ACLR) and healthy controls, report TTDPM is reduced in ACLD and ACLR knees, but no difference in joint position sense between any groups. One would expect there to be consistent differences in proprioception between subjects that have suffered ACL injury and the healthy population if the ACL functions as a proprioceptor, however, as Pap et al. have discussed, input from the ACL and other capsular and ligamentous structures may provide a different proportional amount of afferent feedback from the knee. This may help to explain contradictory results of the investigation of knee proprioception. The role of proprioceptors in the ACL during normal daily activities, in which small loads are applied to the knee and critical end ranges of knee motion are unlikely to occur, may be less consequential then contributions made during strenuous dynamic activity. Rozzi et al.[112] suggest that excessive joint laxity of women appears to contribute to diminished joint proprioception, making the knee less sensitive to potentially damaging forces, therefore increasing the possibly of risk for injury. However the relationship between laxity and proprioception is still unclear. The extent and means to which proprioceptive input from the ACL contributes to knee stability in high demand activities warrants further quantification.

Skeletal dimension, as it is related to lower extremity alignment, is another intrinsic factor entertained as a possible predisposing factor for ACL injury. Specifically, the quadriceps (Q) angle, which is created by two imaginary lines: 1.) from the anterior superior iliac spine to the center of the patella, and 2.) from the center of the patella to the tibial tuberosity, is a skeletally based measure useful in evaluating lower extremity alignment. Females generally have a larger Q angle than males, although skeletal contributions to this difference are not clearly understood. The assumption that females have wider pelves then males has been discounted regardless of whether width is expressed as a measure of biiliocrital, bitrochanteric, or ASIS-ASIS breadths.[18] Further attempts to credit the difference to be the result of pelvic width and femur length have resulted in poor correlations to Q angle.[18, 58] Byl et al.[18] suggest consideration of the fact that the patella, which is mobile and therefore responsive to muscular contraction, be considered as a potential contributor to variations in Q angle. These authors reported a weak but significant relationship between knee extensor torque and Q-angle. This outcome lends potentially greater usefulness because Q angle was measured in standing. Since an increase in Q angle may be related to an increase in knee valgus, determining Q angle in a more functional position, as opposed to a supine non-weight bearing position, offers a more useful estimate of lower extremity alignment. Although Q angle is determined from two rigid landmarks on the pelvis and tibia, the position of the patella, while relying on quadriceps tension, will secondarily depend upon the angle of tension produced by the quadriceps. Frontal plane alignment of the femur logically relies on hip abductor strength or tension. At this time, there are no reports on the relationship of hip abductor strength and Q angle. Further identification of hip abductor weakness as a contributor to lower limb alignment offers potential modification through identifying weakness and implementing appropriate intervention. Therefore, investigation of Q angle in a standing position in addition to hip abductor strength with comparison among males and females is warranted.

Sex hormones have been considered as having an influence on ACL strength and stiffness and are therefore implicated as a potential risk factor for ACL injuries, particularly in females. During the course of the menstrual cycle concentration of endogenous circulating hormones fluctuate considerably, namely estrogen and progesterone. The demonstrated presence of receptors on the ACL for estrogen and progesterone suggests that ligament structure and function may be affected by circulating hormones.[75] Researchers have determined ACL cellular metabolism is decreased in the presence of increased estrogen levels.[132] Specifically, fibroblast proliferation and procollagen synthesis (Type I) express a dose dependent acute reduction, resulting in diminished strength of the ACL for a period of 3-7 days. Progesterone appears to attenuate estrogen effects, especially in the presence of reduced estrogen concentrations.[133]

The normal 28-day menstrual cycle is characterized by two distinct phases in addition to menses. The follicular phase represents the period from the onset of menses until ovulation occurs (day 14 ± 2). The luteal phase starts immediately after ovulation and last for 14 days.[6] Concentrations of estrogen and progesterone are lowest during menses. Estrogen concentration reaches peak just before ovulation, drops sharply, to rise again with a broad peak during the luteal phase. Progesterone concentration rises gradually, beginning just before ovulation, also reaching a broad peak in the luteal phase.[120]

The effect of estrogen and progesterone on knee joint laxity has received recent increased attention. However, results are contradictory. Heitz et al. reported an direct relationship between the concentration of estrogen and progesterone with ACL laxity for females.[50] Assessing hormone plasma concentrations through blood sample and ACL laxity with the KT-2000 knee arthrometer at three phases of the menstrual cycle, the authors reported greatest laxity during the luteal phase. Additionally ACL laxity increased in conjunction with increased estrogen levels from baseline during the follicular phase and with increased progesterone levels from baseline during the luteal phase. No significant differences in knee joint laxity were reported for males across a similar time period. Shultz et al.[116] demonstrated estradiol (the principle estrogen produced by the follicular cells of the ovary), progesterone, and testosterone concentrations contribute to changes in knee laxity across the menstrual cycle. Interestingly, this relationship was stronger when hormone concentrations are compared to knee joint laxity 3-4 days later, suggesting there is a time shift between hormone fluctuation and laxity. ACL stiffness has also been reported to correlate to estradiol concentrations in active females, decreasing during periods near ovulation[30, 111] and the luteal phase.[30]

Conversely, there is an equally persuasive amount of information demonstrating that knee joint laxity does not vary during the menstrual cycle.[7, 68, 123] Most recently, Beynnon et al.[11] reported that although females have greater knee laxity compared to males, overall laxity does not change over a normal menstrual cycle, indicating no relationship between estradiol and progesterone fluctuation and laxity measurements. The authors offer their findings may differ from contrasting reports demonstrating a positive relationship and knee joint laxity due to differences in serum hormone concentrations in subjects in the different studies. Also, although repeated measurements of knee laxity were found to be slightly different, the variability of the repeated measurements may be large enough to prevent evidence of significance. Due to the lack of agreement with regards to hormone fluctuation and hormone concentration further investigations are needed.

Testosterone has received much less attention with regards fluctuation and concentration across the menstrual cycle. Shultz et al.[116] reported small but significant fluctuations in testosterone levels in females across the menstrual cycle. Although testosterone levels are known to vary across women according to the nature of physiological stress, this finding was unique and invited further investigation. Subsequently Lovering and Romani[80] determined the presence of androgen receptors on the female ACL but found although the ACL is an androgen-responsive tissue testosterone is not an independent predictor of ACL stiffness near ovulation.

To date, there is a dearth of information available with regards to the effect of menstrual cycle on athletic performance. In a review of the literature, Lebrun[72] suggests that since the possibility that some physiological responses may change during the menstrual cycle, researchers using females as subjects should standardize the phase of menstrual cycle in which they are tested. Cycle-phase detriment to performance has been reported retrospectively in 37-64% of females, whereas 13-29% report improvement during menstruation.[71] Best athletic performances occurred most often in immediate postmenstrual days, with the worse performances during the premenstrual interval and the first few days of menstrual flow. Survey data provide subjective quantification of the relationship between menstrual cycle and performance, however, lack objective measurement of menstrual cycle phase and endogenous hormone concentrations. Additionally, the term performance is ambiguous and represents several possible indicators (e.g. muscular strength, muscular endurance, aerobic capacity). Sarwar et al.[114], in 1996, determined that quadriceps and handgrip voluntary isometric force increased at mid-cycle compared to the follicular and luteal phases in females not taking oral contraceptives, and suggest increases in estrogen prior to ovulation contribute to these differences. Posthuma et al.[109] reported fine motor function is better during the late luteal phase than the follicular phase in females having asymptomatic premenstrual symptoms compared to those that are symptomatic. Hewett[51] suggests estrogen may have significant effects on neuromuscular control for female athletes.

ACL injury rates have been reported to be increased at different phases of the menstrual cycle. In an investigation of team handball players, Myklebust et al.[101] reported a majority of ACL injuries occurred during the late luteal phase of females, with the second highest incidence during menses, suggesting there may be an increased risk of ACL injury during the week prior to or after the start of the menstrual period. Slauterbeck et al.[120] reported a majority of females suffered ACL injury during the follicular phase and 27% of these injuries occurred immediately prior to and during the first 1-2 days of menses. However, Wojtys et al.[128] reported more injuries around ovulation and significantly fewer injuries in the follicular and luteal phases. In light of the results of reported injury rates and cyclical fluctuation in endogenous hormones and knee laxity, there is a lack of determination of the relationship between hormonal fluctuations and injury rates.

Although increased ACL laxity appears to be concomitant with increased estrogen concentration, the available information with regards to ACL injury rates for females does not conclusively support increased injury incidence during times of the menstrual cycle when estrogen concentration is highest. Part of the confounding results may be due in part to the methods used to determine the phase of menstrual cycle in which the injuries occur, specifically self reporting, and the small sample sizes reported in these studies. Researchers have yet to determine the relationship between ligament laxity and injury risk, regardless of the effect of gender. Investigations which compare neuromuscular and biomechanical variables across different phases of the menstrual cycle are warranted. Until further information is available, researchers should acknowledge the potential effect of menstrual cycle phase on knee joint laxity and neuromuscular control and standardize the phase of cycle in which female subjects are tested in neuromuscular and biomechanical investigations.

ACL injury can occur due to contact with an opponent but predominantly ACL injury is due to non-contact mechanisms, whereby the ligament fails while performing locomotive movements. Non-contact ACL injury reportedly accounts for 70-78% of all ACL ruptures in the athletic population.[15, 97] Non-contact ACL mechanism of injury, in sports such as basketball, soccer, volleyball and football is most often associated with rapid deceleration.[15] Rapid deceleration commonly occurs in three situations: 1.) double or single leg landing from a jump[45, 97], 2.) stopping suddenly when running [45, 97], and 3.) redirecting the path of travel with a plant to cut and run technique.[3, 15, 26]

Jump landings and sudden or abrupt stops are associated with either a double or single leg landing. Ireland [63], based upon video analysis of several non-contact female knee injuries, identified these types of stops to be rapid and awkward, developed in the presence of forward momentum, and demonstrated with consistent postural constraints, such as an upright trunk and limited knee flexion. The timing of the injury is believed to occur quickly after initial foot contact, when the knee is closer to full extension.[15, 103] In this position abnormal internal/external rotation and valgus positions of the knee are potentially more harmful to the ACL.[85] Additionally, positions of the foot, hip and trunk, relative to the knee, create complex segmental relationships which are believed to more likely produce significant ACL strain, making the knee victim to the distal and proximal segmental positions and muscle activity.[63]

Awkward landings are coupled with increased foot pronation, external tibial rotation, internal femoral rotation, and limited hip flexion at the time of the abrupt deceleration.[63] This landing posture is compounded by a position of hip abduction[63], offsetting the center of mass from the knee, resulting in knee valgus buckling. These landings have the intended goal of controlling the center of mass and maintaining balance over the base of support but in the absence of appropriate segmental alignment and muscle contribution to dynamic stability the ligaments may be disproportionately required to resist loads in out of plane positions.

Often the rules and environmental restrictions of the sport being played dictate what athletes intend to do upon landing. For example, a basketball player landing after obtaining a rebound will likely attempt to quickly jump again to shoot the ball or maintain control when landing to avoid taking extra steps and look to pass the ball to a teammate or dribble the ball. A receiver in football attempting to catch a ball near the sideline will attempt to land as to obtain control of the ball in bounds. A front row volleyball player who has hit the ball will be restricted by the need to avoid net contact and the need to read the defense return of the ball and reposition quickly to jump again for a block. Contextually, all of the described scenarios can result in an awkward, out of plane landing.

Rapid changes in direction while running also have potential to create extreme out of plane joint positions, however, there is less concise information available with regards to this mechanism of injury for the ACL. Self reporting by athletes and observations by clinicians[5, 15, 97] have resulted in rapid changes in direction being categorized as a cutting maneuver. The predominant research paradigm for a cutting maneuver involves straight ahead running abruptly redirected at an angle of 30-90 degrees from the original direction of travel.[3, 8, 10, 24, 26]. [82, 90, 93, 94] This maneuver has been termed sidestep cutting and the potential for ACL disruption is theorized to occur due to a combination of frontal, sagittal, and transverse plane knee positions and loads. Although a combination of stresses on the knee result during typical sidestep cutting, research to substantiate whether the resultant stresses are significant enough to disrupt the ACL and which combination of stresses are the most deleterious to the ACL, remain to be determined. Currently, the focus is on resulting knee valgus and rotation and the concomitant hip adduction and rotation in addition to foot pronation during sidestep cutting.[91, 92]

At this time it is unclear if the detrimental knee position during sidestep cutting and jump landings are identical, therefore further investigation of the sidestep cutting mechanism is warranted. Increased hip abduction is not typically found during traditional sidestep cutting techniques. However, current sidestep cutting studies do not allow for a position of increased hip abduction since subjects are required to perform a pure form of sidestep cutting maneuver without any other 'pre-cutting' alterations in performance. Whether or not a position of increased hip abduction is preventative to knee valgus, or can increase the consequence of abnormal knee valgus is not clear. This may help explain the differences between proposed mechanisms of injury for jump landing and cutting conditions.

The purpose of the performing a sidestep cut during running is to misdirect or avoid an opponent. As with landings, the athlete must control the center of mass over the base of support in order to maintain balance during the evasive maneuver. However, this maneuver does not demarcate the end of a locomotor skill as clearly as a jump landing, rather it is a skill used to transition between two directions of travel. The distinct beginning and ending of running direction are less clearly defined. In order to produce an effective evasive maneuver the athlete must: 1.) perform the sidestep cutting maneuver powerfully to accelerate past the opponent and gain an advantage or 2.) increase the deception associated with the maneuver, thereby creating a response by the opponent which is increases the likelihood the athlete will gain an advantage into available space. Several options are available to deceive an opponent. Brief hesitations in running direction or a false step into a direction different then the one the athlete intends to run are commonly performed during sports that require rapid changes in direction.

The performance of a rapid change of direction while running must also be considered with regard to the sporting environment. A running back in football has a primary goal to avoid contact by an opponent, overall making progress with the ball toward the goal line. He may try to evade his opponent by sequencing short runs lateral and diagonal to the desired direction and transition from running in one direction to another with a powerful plant and cut maneuver. Typically he runs lower, generating power from the lower extremity to protect him from imminent contact and maintain possession of the ball and obtain forward progress. A wide receiver will change direction as part of a pre-determined pass route and once he obtains possession of the ball may adopt a similar running style of the running back, depending upon the area of field and relationship to the other defenders. Successful execution of the pass route depends in part on his ability to execute the change of direction in the route to gain advantage over his defender. Therefore, the football player must successfully change direction with and without possession of the ball.

Rapid changes with and without the ball occur in basketball and soccer as well, although the most obvious difference is that neither of these sports allows individuals to hold onto the ball and run. These athletes must maintain control of the ball with coordinated dexterity of the hands or feet. Basketball and soccer players who are dribbling will utilize evasive maneuvers, at the same time maintaining and protecting the ball from the opponent. Logically, sequencing of hesitations and false steps in a variety of directions, while overall desiring to bypass the defender and protect the ball necessitate a different overall rhythm without the consequence of receiving direct contact like the sport of football.

An offensive basketball player is not as concerned with avoiding or preparing for contact, but rather utilizes quick deceptive movements to potentially place the defender in a position of disadvantage so the ball handler can move into more freely available space. Although pure side step cutting may be utilized to gain this advantage, it is unlikely that it occurs in isolation without a combination of a brief hesitation or false step into a direction different then the one the athlete drives into while dribbling. The basketball defender will spend a highly concentrated time moving laterally and or running forward to recover position and will attempt to avoid ensuing contact of screens set by opponents. The planting and cutting maneuver will most likely occur when transition is made from a lateral shuffle movement to a sprint or backpedal or vise versa or when the screen is anticipated. It is more likely that the defender will decelerate more suddenly while responding to the offensive player or ball movement for both the sports of basketball and soccer. The basketball offensive player will decelerate more suddenly to maintain ball control, avoid traveling, or stay in bounds while maintaining possession of the ball.

The constraints of the playing area are greater for basketball than soccer and may contribute to the frequency and suddenness in which rapid changes occur, although this has not been substantiated in the literature. There is a paucity of qualitative evidence of the specific types of movements, frequency of occurrence and speeds at which these tasks are performed in sports that require rapid changes of direction.

Cutting maneuvers are not typically performed at maximum running speeds.[93] It has been proposed that common running speeds in which run and cut maneuvers are performed range from 5.5 to 7.0 $m \cdot s^{-1}$.[93] It is not known whether these speeds are representative of all types of sports, athletes, and male and female athletes. Anecdotally researchers agree that jumping, cutting, shuffling, backpedaling and running are typical. In an investigation of professional male basketball players, McClay-Davis et al.[88] found movements common to the sport (running, cutting, starting and stopping, lay-up take-off and landing, jump shot take off and landing, vertical jump take off and landing and shuffling) result in different knee and ankle positions and joint velocities. Knee flexion was relatively similar for all tasks with the exception of starting, vertical jump take-off, and shuffling. These movements required larger knee flexion angles on average. Stopping and shuffling tasks had larger maximum knee extension, suggesting the knee is relatively more flexed throughout the movement. Knee flexion velocity was found to be higher for activities such jump landings and lay-ups. Despite the information provided by this study, there is a considerable lack of information available on these similar characteristics for other sports, for example soccer and football. Characterization of the typical frequency, sequencing, speed, and distances over which these movements are performed, and secondly if differences exist across sports, would likely improve quantifying exposure rates to potentially injurious movement patterns.

Researchers have adopted the term neuromuscular control to identify mechanisms and strategies for enhancing joint stability and positioning during movement tasks. Griffin et al.[45] state neuromuscular control refers to the “unconscious activation of dynamic restrains surround a joint in response to sensory stimuli”. Neuromuscular control is a component of the broader area of motor control. Motor control is the ability to regulate or direct the mechanisms essential to movement.[117] Motor control relies on an interaction of the individual, task and the environment through cooperative action of the peripheral and central nervous systems and muscular systems. Therefore, an argument can be made that conscious activation of dynamic restraints surrounding a joint also play a role in neuromuscular control. Behavior can be learned and trained to be appropriate to the specific task and relative to the environment in which it is required. Perception, which is the integration of sensory impressions into psychologically meaningful information, and cognition, which includes attention, motivation, and emotional aspects of motor control, underlie our intent and ability to act effectively in the environment.[117] Understanding the role that neuromuscular control has in the prevention of knee ligament injuries and the consequences of impaired neuromuscular control subsequent to injury has led to intense interest in the study of this area of research related to ACL injuries.

Neuromuscular control is related to several important variables: 1.) proprioceptive feedback, 2.) short and long latency reflex responses, 3.) synergistic muscle recruitment, 4.) feedforward mechanisms for muscle activation, and 5.) muscular strength. The interplay amongst these variables directly affects the timing and recruitment patterns of muscular activation, providing coordinated and appropriate muscle activity to control position during locomotive skill performance. Knee joint stability is dependent upon passive and dynamic restraints. As already discussed, key passive restraint to anterior translation of the tibia is provided by the ACL. Knee positions during activity, which result in increased loads on the ACL, can be managed dynamically through activation of muscles that act directly to control knee position and indirectly to control foot, tibia and femur position at the ankle and hip.

Three of the four quadriceps muscles originate directly on the femur, whereas the rectus femoris, a biarticulate muscle, originates on the pelvis. All four form a common insertion onto the tibial tuberosity. Therefore, a great capacity to control tibia and femur position is possible in the sagittal plane. The hamstrings however, have a directly opposite relationship to the femur and pelvis as only one of the hamstring muscles originates on the femur and the remaining three originate off of the pelvis. Three distinct insertions of the hamstrings onto the tibia and fibula on the medial and lateral aspect of the knee lend to the capacity of the hamstrings to individually help control transverse position of the leg, and collectively, position of the tibia relative to the femur in the sagittal plane. The hamstrings originating on the ischial tuberosity of the pelvis are more likely to affect pelvis position than the single quadriceps muscle (rectus femoris) which originates on the anterior superior iliac spine. The semitendinosis, semimembranosis and long head of the bicep femoris are secondary hip extenders in addition to primary knee flexors. Therefore, they contribute to femur position through their primary contributions to pelvis and tibia motions. Position of the femur in the frontal plane is primarily controlled by the hip abductors and hip adductors. The gluteus medius, acting from the iliac crest to the greater trochanter of the femur, is a primary hip abductor and assisted by the tensor fascia latae insertion into the iliotibial tract, which inserts on the lateral aspect of the knee. Hip adduction is created by several large adductor muscles which have origin from the pelvis tubercle and insertion on the femur in addition to the biarticulate gracilis muscle. The gastrocnemius muscle, also a biarticulate muscle, crossing both the knee and the ankle, contributes to knee flexion, ankle plantar flexion, and internal and external rotation of the femur relative to the tibia.[113] Norkin and Lavangie[102] suggest the gastrocnemius contributes more minimally to knee flexion than ankle plantar flexion, and dynamically stabilizes the knee by resisting large extension torque at the knee during normal gait.

Although most muscle function is defined relative to origin and insertion across specific joints, realistically, synergistic action of muscles in planes other then normally identified is important for neuromuscular control. Lloyd and Buchanan[79] found hamstring and quadriceps co-contraction provided significant support in conditions of increased valgus and varus loads on the knee, in addition to the gracilis and tensor fascia latae. Similarly, Besier et al.[8] suggest co-contraction of the hamstrings and quadriceps help to counter external valgus loads at the knee. Therefore, it seems evident that consideration of muscle activity during athletic maneuvers should not be relegated to the prime movers for particular movements, but rather globally to lower extremity stability and function. Muscle synergies have long been identified to help control postural perturbation, therefore determining if specific patterns of muscle activation are beneficial to the control of lower extremity position and reduction of hazardous loads on the knee is important in the study of ACL injury.

Most of the research related to muscle activity at the knee has focused on the strength and activation patterns of the quadriceps and hamstring muscles. Huston and Wojtys[61] and Anderson et al.[2] report male athletes have greater quadriceps and hamstring strength, even when adjusting for body weight, than female athletes. Primarily knee extenders, the quadriceps are considered ACL antagonists since isolated activation of the quadriceps at knee flexion angles from 0-80° produces significant anterior tibial translation.[56] Anterior translation is greater between 0-30° of knee flexion compared to positions between 30-90° of knee flexion with isolated quadriceps load. [74] Peak anterior translation reportedly occurs at 30° of knee flexion.[56, 74] Primarily knee flexors, the hamstrings are considered agonists to the ACL, since activation of these muscles results in posterior translation of the tibia. Hamstring activation is more effective in producing posterior translation of the tibia at knee flexion angles greater then 15° because these muscles have a relatively low mechanical advantage between 0-15° of knee flexion.[74] Furthermore, Pandy and Shelbourne[104] state co-contraction of the hamstrings and quadriceps cannot protect the ACL near full knee extension because the hamstrings meet the tibia at smaller angles and cannot apply a sufficient posterior shear force to the tibia. Consequently, ACL strain is reduced with co-contraction of the hamstrings and quadriceps, particularly at angles greater then 15° of knee flexion.[13] Isolated quadriceps load eventually results in posterior translation at 120° of knee flexion.[74] This is due to the change in the direction of pull of the quadriceps relative to the tibia at greater knee flexion angles. Therefore, the relative strength of the quadriceps and hamstrings play an important role in dynamic stability of the knee and inadequate strength may result in less protection available for the ACL, particularly at knee positions closer to full extension.

Huston and Wojtys[61] determined males and females demonstrate different activation patterns of the quadriceps and hamstrings subsequent to anterior tibial perturbation. In a position of 30° knee flexion, female athletes relied more on the quadriceps and gastrocnemius muscles to resist anterior tibial translation initially, whereas males more frequently recruited the hamstrings to resist anterior tibial translation. This appears counterintuitive since the quadriceps is antagonistic to the ACL at a position of 30° of knee flexion. Although the non-contact mechanism of ACL injury does not include mechanical perturbation of the tibia, the results are none-the-less interesting. Hewett et al.[52, 55] suggest that females tend to demonstrate a quadriceps dominant recruitment pattern when performing activities such as jump landings. This pattern is believed to place more strain on the ACL. Whether or not this phenomenon is due to habitual quadriceps dominant recruitment patterns or preexisting imbalanced quadriceps to hamstring strength ratios has yet to be determined. Integrated EMG (IEMG) of the quadriceps during cross over cutting and sidestep cutting tasks was found to be greater for females then males by Malinzak et al.[82] This 17-40% increase in quadriceps IEMG was concomitant with an approximate 20% reduction of hamstring activity across stance. One of the strongest contentions by these researchers is neuromuscular training successfully reduces imbalances in quadriceps to hamstring strength ratios and diminish the quadriceps dominant recruitment pattern displayed by females. Evidence of the effects of training on improving knee stability through increased strength and appropriate recruitment patterns of the quadriceps and hamstrings deems neuromuscular control as an external variable associated with knee injury and therefore mutable.[51]

Contrary to this evidence, other researchers have reported no difference in muscle recruitment patterns between males and females on selected athletic tasks. Fagenbaum and Darling[34] found no differences in hamstring and quadriceps activation immediately prior to and during jump landing tasks for female and male basketball players. Colby et al.[24] reported that hamstring activity increases prior to foot contact for sidestep cutting and cross over cutting maneuvers but diminishes or remains constant after foot contact. Quadriceps muscle activation begins just before foot strike and peaks in the middle of eccentric motion during cutting, stopping and landing tasks. Although these authors studied males and females, comparison was not performed; therefore, muscle activation patterns for these selected tasks, was descriptive of both males and females. Likewise, Besier et al.[8] found average EMG of 10 muscles of the lower extremity to be greater during cutting tasks then running for males. Evidence of selective co-activation strategies to stabilize the knee in varus-valgus and internal-external rotation directions when the cutting task was pre-planned as opposed to a generalized co-contraction of muscles during unanticipated cutting maneuvers, suggests pre-planned activation patterns better control loading of the knee in those directions. While is has been accepted sagittal muscle activation patterns control flexion-extension directions of the knee, this was one of the first studies to demonstrate control in the frontal and transverse planes.

Despite investigations of muscle activity during jump landing and cutting tasks, there is inconclusive evidence of a gender effect on activation patterns. It can be accepted activation strategies are different across selected athletic tasks and excessive quadriceps torque can be detrimental to the ACL. However, the ability of quadriceps activation to be large enough to result in rupture of the ACL during these tasks is controversial. Theoretically, because landing and cutting maneuvers will generate large quadriceps forces, especially in positions of smaller knee flexion angles, anterior shear force on the tibia will be increased.

In an investigation of sidestep cutting, McLean et al.[90] reported that sagittal plane knee joint forces are not large enough to rupture the anterior cruciate ligament. Through the use of Monte Carlo simulations to predict peak anterior drawer force at the knee during sidestep cutting maneuvers, values never exceeded 900N, well below 2160N, reported to be the injury threshold for the ACL in young adults.[130] Their model included situations in which quadriceps force was nearly doubled and hamstring force was zero. These results are in opposition to other researchers who propose that sagittal plane mechanisms can account for ACL injury in the presence of a powerful quadriceps muscle force.[24, 104] The effect of sagittal plane muscle activation strategies and resulting loads on the knee cannot be ignored. McLean et al.[90] advise combinations of resulting loads placed on the knee in the sagittal, frontal, and transverse planes may be required to injure the ACL. Determination of which muscle activation patterns in the entire lower extremity reduce or produce detrimental ACL loads and if gender differences exist, may offer a better explanation of the mechanism of ACL injury in landing and cutting tasks.

Less information is available with regards to the effect of hip strength and it's correlation to knee position during athletic tasks. Recently, in an investigation of eccentric hip abductor strength and knee joint kinematics during jump landing tasks, Jacobs and Mattacola[64] determined that although there were no differences between male and female peak eccentric hip abduction strength (normalized to body weight), there was a significant inverse relationship between female's peak eccentric strength and peak knee valgus angle. Specifically, females demonstrating larger eccentric peak torques were also found to have lower peak knee valgus angles during landings. These results are interesting; especially in light of the fact hip abductors have the capacity to control femur position in the frontal plane and excessive frontal plane valgus of the knee. Frontal plane knee valgus, the result of femur and tibial position, is theorized to contribute to ACL injury. Currently there are no investigations which report a correlation between hip abductor strength and frontal plane position of the knee during cutting tasks, or an effect of gender. This invites examination of these relationships in order to help understand how hip strength is related to lower extremity alignment during dynamic tasks involving rapid changes of direction.

Biomechanical studies provide information about kinematics (the form, pattern, or sequencing of movement with respect to time) and kinetics (the relationship between forces acting on a system and the motion of the system). Advancements in technology have resulted in increased ease of collection and analysis of kinematic and kinetic data; therefore, there has been a recent surge of studies focused on the analysis of athletic movements. In particular, investigations focused on better describing joint position and loads during landing and cutting maneuvers have provided objective information related to mechanisms of ACL injury.

Males and females demonstrate different lower extremity mechanics in the transverse and frontal planes during running.[35] Ferber et al. [35] found female recreational runners have significantly greater peak hip adduction, hip internal rotation and knee abduction angle compared to males. Therefore, it seems apparent that gender differences exist even in cyclical locomotor activities and plausible that differences will be evident during other athletic tasks. Early attempts to quantify the biomechanics of jump landings, sudden stopping, and rapid changes in direction have centered on identifying relative knee position and moments in the sagittal, frontal and transverse planes during these tasks. As previously identified, these three maneuvers are believed to be primary occasions in which non-contact ACL injury occurs. Anecdotal evidence of awkward landings being associated with limited knee flexion, foot pronation, tibial rotation, femoral rotation, and varying amounts of hip flexion at the time of the abrupt deceleration[63] have spurned the generation of models to explain how these positions and relationships can be detrimental to the ACL. As such, current investigations not only focus on knee biomechanics, but the foot, ankle, hip and trunk as well. What is less clear, is whether the same segmental orientations during jump landing and sidestep cutting result in ACL injury?

Researchers have proposed one of the contributing factors to the differences in ACL injury rates between males and females is due to the belief females generally perform athletic tasks with less knee flexion than males. Realizing there is less protection afforded the ACL due to co-contraction of the hamstrings and quadriceps of knee flexion angles less than 15°, this idea is inviting. However, there is conflicting evidence of females demonstrating a knee extension bias during athletic tasks. Colby et al.[24]compared the results of average knee flexion at foot strike in four different tasks: sidestep cut (22°), cross over cut (29°), stopping (14°) and jump landing (23°). Although the authors did not report the results of statistical comparisons of these knee flexion values, average knee flexion values (22°) for all of the tasks are similar to the knee position reported by athletes that could recollect the approximate knee flexion angle at the time of ACL injury.s[97] No comparison of males and females knee flexion angles was reported in these two studies.

Peak knee flexion angles have been found to be greater during lateral false step maneuvers compared to running in females, although a gender comparison was not available.[43] In an investigation of knee joint mechanics during sidestep cutting for males, both McLean et al.[93] and Besier et al.[9] determined peak knee flexion angles to be greater during sidestep cutting than running. Subsequently, McLean et al.[94] reported that although the average range of motion for females and males was not different during sidestep cutting tasks, females reached peak knee flexion earlier across stance than males. However in another investigation, McLean et al.[90] suggest females have less knee extension and hip extension at landing, limiting optimal control provided by sagittal plane muscles. Malinzak et al.[82] reported that knee flexion angle was generally less during running, sidestep cutting and cross over cutting maneuvers for females than males. In an investigation of jump landing strategies for male and female collegiate athletes, Fagenbaum and Darling[34] found females land with 10-14° more knee flexion than males. An increase in knee flexion during athletic tasks is deemed protective of the ACL. When individuals rapidly decelerate body mass in the sagittal plane, compensation for increased loading on the knee will occur by increasing knee flexion.[93] Further investigation of the differences in knee flexion angles between males and females during lateral false step and sidestep cutting tasks is warranted.

Interest in frontal plane knee position is currently at the forefront of investigations related to jump landings and sidestep cutting maneuvers. Hewett[51] indicates that landing from a jump with either a valgus or varus lower extremity alignment creates a less stable position for the knee and further suggests that knee valgus is a predictor of ACL injury for females.[54] In 1999, McLean et al.[94] determined the knees of females to be more abducted near foot contact and remain more abducted throughout stance during sidestep cutting maneuvers than males. McLean et al. have also reported females perform sidestep cutting maneuvers in the presence of a defender with greater peak knee valgus and rearfoot pronation than males.[92] Although the peak difference in peak valgus between males and females was 2° in this study, the authors determined this would lead to a 40Nm increase in the valgus moment; equating to an increase of 100%. These authors' further report females sidestep cut with less peak hip flexion, hip abduction, and hip internal rotation than males. These results are in agreement with Malinzak et al.[82] who reported that females have greater knee valgus across stance during sidestep cutting than males. However, other research has found no differences in peak knee valgus across stance during the first 0-40° of knee flexion.[107] Pollard et al.[107] tested male and female collegiate soccer players in conditions in which the cutting maneuver was performed in a randomized manner. Order of either cutting, straight ahead run, and jump stop were performed randomly in response to a visual signal. The results of knee valgus angles were determined from cutting trials only.

There is considerably less information available with regards to hip frontal plane position during athletic tasks. Pollard et al.[107] found females demonstrate significantly less peak hip abduction during sidestep cutting than males, which is in agreement with McLean et al.[92] Both authors investigated sidestep cutting in conditions where the subjects responded to external stimuli, Pollard et al. with visual signal related to performance of different tasks, and McLean et al. with the presence of defensive opponent. The relationship between hip abduction and knee abduction is interesting and the significance of this information may be helpful in explaining knee valgus. Therefore, consideration of hip position during athletic maneuvers warrants further investigation.

Transverse plane motion at the knee during athletic maneuvers may also contribute to ACL injury. As previously indicated, during rapid stopping and jump landings, the mechanism of ACL injury is believed to be associated with external tibial rotation. Although the ACL functionally provides restraint to internal rotation, Markolf et al.[85] measured increases in ACL force with externally applied external rotation at the knee with cadaver specimens. ACL force is greater during external tibial rotation in the presence of quadriceps force and increases at angles greater than 0° of knee flexion. Internal rotation produces greater ACL force than external rotation at all positions of knee flexion, but does not appear to be affected by the addition of quadriceps force. The possible explanation for external rotation contributing to ACL strain may be due to the change in orientation of the ACL relative to the intercondylar notch, i.e. an impingement pattern. Model simulations have been used to explain ACL impingement during positions of knee abduction and external rotation, particularly in the presence of a narrow intercondylar notch.[42] Therefore, although not intuitive based upon the anatomical orientation of the ACL in the knee, it appears that external rotation in combination with knee valgus has the potential to injure the ACL.

McLean et al.[93] found sidestep cutting results in greater tibal external rotation when compared to running, subsequently reporting females display less internal rotation than males for this maneuver.[92] These authors report no period of absolute internal rotation during sidestep cutting. Conversely, authors have reported no significant difference in peak knee internal rotation between males and females.[107] Altering running stride by increasing the distance the limb makes contact with the ground in the lateral direction and then redirecting back into the original path of travel significantly increases knee internal rotation compared to running.[43] Therefore it appears sidestep cutting and lateral false step maneuvers have different effects on transverse plane motion at the knee, both potentially hazardous to the ACL.

Few studies have reported hip rotation positions during athletic maneuvers. Available results are contradictory as researchers have reported no significant differences in peak hip internal rotation between males and females[107], whereas others have found hip internal rotation to be greater for females then males.[90] Interestingly, McLean et al.[91] found knee valgus moments to be correlated to hip rotation, in addition to hip flexion and knee valgus. This lends support to the idea the entire lower extremity should be under consideration in helping to explain differences in knee position and possible mechanisms of injury.

Jump landings and sidestepping are expected to produce significant moments (loads) on the knee than activities of daily living such as walking and running. The consequence of these different loads on ligaments and other soft tissues may be significant enough to produce injury. Hewett et al.[55] state reducing valgus and varus torques at the knee and hip during landing potentially aid in stabilizing the knee joint and preventing injury. It appears females who demonstrate large abduction loads during landing are at increased risk of ACL injury.[54] In a prospective study including 205 female athletes, Hewett et al.[54] found athletes demonstrating dynamic valgus of 8.4° or greater and peak knee valgus moments of 27 Nm or greater were more likely to injure their ACL during their sport season.

Frontal and transverse plane moments have been reported to be greater during sidestep cutting at 30 and 60°[10] and lateral false step maneuvers[43] compared to running. There is conflicting evidence of differences in frontal plane moments between males and females. Researchers have reported peak external knee valgus moments to be significantly greater for females than males for sidestep cutting.[91] Conversely, Pollard et al.[107] found no difference in peak internal knee adduction moments between males and females. These authors also reported no differences in internal hip abduction and external rotation or knee external rotation moments between males and females.

Sagittal plane moments, specifically flexion, although larger than frontal and transverse plane moments, for both running and sidestep cutting, do not appear to be significantly different between tasks.[10] Similarly, lateral false step maneuvers do not appreciably affect sagittal plane moments at the knee.[43] The contention that sagittal plane forces are deleterious to the ACL because large quadriceps forces are evident in positions of relatively smaller knee flexion angles has been used to explain ACL mechanisms of injury.

Based upon a modeling estimation of peak anterior drawer force during sidestep cutting, McLean et al.[90] propose forces never reach the critical value of 2000N in any simulated conditions. This is not to suggest loading in the sagittal plane cannot in part help explain ACL injury. Rather, sagittal plane loading in isolation may not explain ACL loading and should be considered in conjunction with loads in the frontal and transverse planes.

In conclusion, knowledge of the potential contributing factors to ACL injury increases the ability of clinicians to recognize which characteristics the athlete possesses. Although certain intrinsic factors, such as ACL dimension and mechanical characteristics, and intercondylar notch width are not routinely and easily quantified in the healthy population, they aid in helping understanding how the ACL functions as a static stabilizer. Limb alignment is typically considered an intrinsic factor and non-modifiable, but identifying the effect of muscular strength on Q angle offers potential for intervention, thus reducing the effect of malalignment as a contributor to injury risk. ACL laxity appears to be related to increased estrogen levels during particular phases of the menstrual cycle for females. Relationship between ACL laxity, at risk phases of the menstrual cycle, and ACL injury have not been clearly delineated, however, consideration of hormone fluctuations and their effect on performance should be entertained during research related to maneuvers which are believed to put the ACL at risk of injury. Although there is strong anecdotal evidence of which conditions results ACL injury during athletics, it is not clear if the alignment of the lower extremity is overwhelming similar to produce injury during jump landings and cutting maneuvers. Both mechanisms appear to rely heavily on the effect of knee valgus. Awareness of which factors increase knee valgus offer great potential in understanding of not only the mechanism of injury but apparent differences in neuromuscular control and biomechanics between males and females. It appears females control of the center of mass over the base of support during landing and cutting tasks through the frontal plane more than through the sagittal plane compared to males. Less muscular support is available to dynamically control and stabilize the segments of the lower extremity in this plane and sagittal plane stabilizers are equally mechanically disadvantaged to offer synergistic control. The resulting solution would be to increase contribution from static stabilizers and consequently increasing risk of injury. In this end, ACL injury is related to several variables. As researchers continue to investigate the structural, hormonal, mechanisms of injury, neuromuscular, and biomechanical factors related to the anterior cruciate ligament, we increase the possibility of predicting which athletes are at risk of injury and what interventions will best serve them, regardless of sport or gender.

APPENDIX B

Date:
ID#:

HEALTH HISTORY QUESTIONNAIRE

Oregon State University
Department of Nutrition and Exercise Sciences

Investigation of Running and Sidestep Cutting Tasks
Demographic and Health History Questionnaire

Please take the time complete this questionnaire thoroughly and honestly by checking the appropriate box and writing in the appropriate information wherever necessary. If you have any questions feel free to ask the investigator for clarification.

PERSONAL

Gender: ❑ Male ❑ Female	Age:	Date of Birth:

SPORT HISTORY

Your primary varsity sport at the collegiate level is ____ soccer, _____basketball?

How many years of competition have you completed at the collegiate level?
□ 1 □ 2 □ 3 □ 4

Are you or did you participate at the Junior College level? □ Y □ N If yes, # seasons?:

How many seasons of organized participation did you complete in your primary sport prior to beginning your college career?

High School	□ 0-1	□ 2	□ 3	□ 4		
Junior High	□ 0-1	□ 2	□ 3			
Club or similar	□ 0-1	□ 2	□ 3	□ 4	□ 5	□ 6 or greater

What other organized sports did you formally participate or compete in during high school?

□ None				
□ Basketball	□ 1	□ 2	□ 3	□ 4
□ Soccer	□ 1	□ 2	□ 3	□ 4
□ Volleyball	□ 1	□ 2	□ 3	□ 4
□ Softball	□ 1	□ 2	□ 3	□ 4
□ Baseball	□ 1	□ 2	□ 3	□ 4
□ Cross Country	□ 1	□ 2	□ 3	□ 4
□ Swimming & Diving	□ 1	□ 2	□ 3	□ 4
□ Track & Field	□ 1	□ 2	□ 3	□ 4

What were your primary events in track and field:

Other sports not listed? Number of seasons?

Health History: Upper Extremity (check the box for the appropriate response)

Are you currently under the care of a physician or athletic trainer for any of the following injuries?	Does this injury limit or prevent you from participating in conditioning drills, practice or competition at this time?
☐ Abdominal Strain	☐ Y ☐ N
☐ Low Back Strain	☐ Y ☐ N
☐ Shoulder Sprain	☐ Y ☐ N
☐ Elbow Sprain	☐ Y ☐ N
☐ Wrist or Hand Sprain	☐ Y ☐ N
☐ Other:	☐ Y ☐ N
Comments:	
☐ Yes ☐ No Do you have any other injury or complication with the upper extremity not listed that you are currently receiving medical treatment for that would hinder your ability to run at maximum speed? If yes, please explain.	

Health History: Lower Extremity (check the box for the appropriate response)

Are you **currently** under the care of a physician or athletic trainer for any of the following injuries? Or have you suffered an injury to these areas that **kept you out of practice for more than three weeks during the six months?**	If yes, does this injury **limit or prevent** you from participating in conditioning drills, practice or competition at this time?	If yes, have you had **surgery** performed to care for this injury or a different injury to this body part? If yes, please explain below in comments section.
☐ Toe or Foot Sprain	☐ Y ☐ N	☐ Y ☐ N
☐ Ankle Sprain	☐ Y ☐ N	☐ Y ☐ N
☐ Achilles Tendonitis	☐ Y ☐ N	☐ Y ☐ N
☐ Calf Strain	☐ Y ☐ N	☐ Y ☐ N
☐ Patella Tendonitis	☐ Y ☐ N	☐ Y ☐ N
☐ Knee Ligament Sprain	☐ Y ☐ N	☐ Y ☐ N
☐ Knee Meniscus Injury	☐ Y ☐ N	☐ Y ☐ N
☐ Thigh Strain: Quadriceps, Hamstring, Groin	☐ Y ☐ N	☐ Y ☐ N
☐ Hip Flexor Strain	☐ Y ☐ N	☐ Y ☐ N
☐ Low Back Strain	☐ Y ☐ N	☐ Y ☐ N
Comments:		
☐ Yes ☐ No Do you have any other injury or complication with the lower extremity not listed that you are currently receiving medical treatment for that would hinder your ability to run at maximum speed? If yes, please explain.		

Menstrual Cycle History: Females Only

□ Yes □ No	Have you had a regular menstrual cycle for the past 3 months? [A regular menstrual cycle means you have had your period (3-7 days in duration) every 28-32 days]
□ Yes □ No	Are you currently taking prescription oral contraceptive medications?
What was the date of the start of your last menstrual cycle?	

APPENDIX C

INFORMED CONSENT DOCUMENT

Project Title: Knee Kinematics and Kinetics as the Result of Combining a Lateral False Step with a Sidestep Cutting Maneuver
Principal Investigator: Mark A. Hoffman, Nutrition and Exercise Sciences
Co-Investigator(s): Grace M. Golden, Michael J. Pavol, Nutrition and Exercise Sciences

WHAT IS THE PURPOSE OF THIS STUDY?

You are being invited to take part in a research study designed to investigate differences in knee responses to different types of running tasks. These tasks include straight ahead running in addition to tasks in which you change direction while running. We believe the positions of the knee and stresses placed on the knee will be different in each of the running tasks. Secondly, we will be measuring hip strength and determining if there is a relationship between hip strength and knee position during running tasks. This research will be useful in helping to understand the cause of knee injuries in athletes. The outcomes of the study will be used for a student dissertation project and results will be submitted for publication and presentation to sports medicine professionals. We are studying this because we want to determine which tasks place the anterior cruciate ligament of the knee at risk for injury and whether or not there are differences between males and females in these tasks.

WHAT IS THE PURPOSE OF THIS FORM?

This consent form gives you the information you will need to help you decide whether to be in the study or not. Please read the form carefully. You may ask any questions about the research, the possible risks and benefits, your rights as a volunteer, and anything else that is not clear. When all of your questions have been answered, you can decide if you want to be in this study or not.

WHY AM I BEING INVITED TO TAKE PART IN THIS STUDY?

You are being invited to take part in this study because you are currently an active healthy high-level male or female athlete who competes at the collegiate level in basketball or soccer. You will be determined to be absent of significant injury to the lower extremities over the past 6 months, due to your completion of the general screening form that you will complete after reading this document. It is anticipated that 15 males and 15 females will participate in this study.

WHAT WILL HAPPEN DURING THIS STUDY AND HOW LONG WILL IT TAKE?

If you agree to participate, your involvement will last for approximately two hours. Only one session is required to complete this study unless due to unforeseen circumstances there are difficulties in administering the testing session. Should that be the case, you are entitled to withdraw from the research study if you do not feel you can volunteer the additional time.

The following procedures are involved in this study. All data collection will take place in the Biomechanics Laboratory and Sports Medicine Research Laboratory, located in the basement of the Women's Building at Oregon State University. Approximate times for each portion of the testing session are included in parenthesis following the description of the particular component of the session.

Part I
- Complete and sign the informed consent and demographic questionnaire. (10min)
- Change in to the appropriate clothing attire: running shoes, compression thigh length shorts and either a sports bra or tank top (female subjects), per your comfort, if you are not already wearing the appropriate clothing. (5min)

- Your height and weight will be measured. (2 minutes)
- You will perform a comfortable five minute warm-up run on a treadmill at a speed of 5.0-7.0 mph. (5 min)
- Small adhesive markers will be place on particular landmarks on your limbs and trunk for camera measurement of your limb lengths and joints during the running tasks. There will be 18 markers placed on your legs and pelvis and 10 on your trunk and arms (including breastbone, left shoulder blade, lower portion of the neck, and low back), 4 on your head (with a headband). (8 min)
- A static trial will be measured. You will stand quietly on a force platform imbedded in the floor in a predetermined position. This allows us to measure your limb lengths prior to performing the running and cutting trials. (2 min)
- You will perform practice trials of regular running and different cutting maneuvers before data collection so you are familiar with the testing procedures. We will be looking for you contact the force plate with your right foot only. Also, we will monitor your running speed and want you to run at approximately 7-8 mph for each trial. This is faster then a jog but slower then a sprint. (5 min)
- Testing will begin, and you will perform 10 successful trials of each of the four different conditions: normal running, lateral false step maneuver, sidestep cutting maneuver, and a combination of a lateral false step maneuver with a sidestep cutting maneuver. A lateral false step maneuver is performed by running straight ahead, then taking a step to the side (widening a step compared to normal running) and then continuing to run forward. The distance of the widened step will be based upon your height. This distance will be 30% of your body height. For example, if you are 5'8" tall, distance you will lengthen your stride is 20 inches. Sidestep cutting involves running straight ahead and then changing your direction by 35-55 degrees from the original direction you were running. The last condition involves combining the lateral false step maneuver with the sidestep cutting maneuver. You will perform the conditions in the order of normal running, lateral false step, side step cutting, and combination of lateral false step with sidestep cutting. You can anticipate that you will complete between 50 to 60 running trials, which are approximately10 yards in length for the entire testing session (500-600 yards total running). You will be able to rest between each trial by walking back to the start location in the laboratory. You will only contact the force plate with the right leg. (40 min)
- When you are done with the running and cutting trials we will remove the markers and escort you to the Sports Medicine Laboratory for the second portion of the session.

Part II

- You will participate in a measurement of your hip strength. This consists of measuring how much force you can produce with the hip abductor muscles. These muscles are located on the outside of your hip. We will be using a machine that controls the speed at which you move and measures how much force you produce during the movement. We will test you at a speed of 120 degrees per second. This is similar to a speed in which you might perform a leg extension exercise during weight training. You will be standing with your right hip aligned with the center of the machine. Your thigh will be affixed to the lever arm of the machine with a padded Velcro fastener. You will perform 2 sets of 5 repetitions. The first set is a practice set. You will have a 2 minute rest period before completing the test set. We will ask you to perform the strength testing with maximal effort. (10 min)
- The final portion of the session involves a measurement of your lower extremity alignment. We will measure your standing quadriceps (Q) angle. This involves placing a plastic goniometer (two plastic lever arms which can rotate about a center fulcrum) on the top of your knee cap. The lever arms are aligned with two lines that are created from landmarks on your leg and pelvis. (10 min)

WHAT ARE THE RISKS OF THIS STUDY?

The possible risks associated with participating in this research project are expected to be minimal. The running, lateral false step, sidestep cutting, and combination of lateral false step and sidestep cutting maneuvers are similar to movements you frequently utilize in your respective sports. Although this research study involves investigation of tasks that may be associated with the mechanism of serious knee injury, the speed of the running trials, distance of the lateral false step and sharpness of the sidestep cutting maneuver have been selected to minimize potentially hazardous stress on the knee but still provide simulation of a movement that has been linked to the mechanism of injury. There is negligible chance of muscular soreness of the legs from the running and cutting maneuvers, since you participate in vigorous exercise 2-3 hrs per day, 3-5 days per week. Eccentric strength testing has the potential to produce short term muscle soreness on the lateral aspect of the hip. Delayed onset muscle soreness from the eccentric strength testing is expected to last 3-5 days from the day of testing. This soreness is similar to what you have experienced when beginning a new exercise program and should have a minimal effect on your ability to continue practicing or training for your sport.

RESEARCH RELATED INJURY

In the event of research related injury, compensation and medical treatment is not provided by Oregon State University.

WHAT ARE THE BENEFITS OF THIS STUDY?

There is no direct personal benefit afforded the participants of this study. The contribution that this study makes in helping to understand knee injury mechanisms, particularly in the sports of basketball and soccer may be of personal interest and satisfaction to you. The researchers anticipate that society may benefit from this study by the contributions made to the current body of knowledge regarding how the knee is affected by rapid changes in direction during running tasks. We hope that, in the future, other people might benefit from this study because better understanding how knee injuries occur may help in the development of intervention programs the decrease the likelihood this injury will occur.

WILL I BE PAID FOR PARTICIPATING?

You will not be paid for being in this research study.

WHO WILL SEE THE INFORMATION I GIVE?

The information you provide during this research study will be kept confidential to the extent permitted by law. It is possible that a portion of these records could contain information that personally identifies you. During the measurement and collection of data, you will only be identified on the computer system with a subject number. The subjects name is to remain confidential in the analysis of any information regarding this study. One aspect of this study involves making video recordings of you when you are running. However, the video cameras used for data collection acquire digital signals from small reflective markers that are affixed to you; therefore, there is no video output that would identify your face or body. Only researchers directly involved in the data collection and analysis will have access to the raw data associated with this study. All information, including health history questionnaires, informed consent documents, and demographic information will be securely stored and accessible only to the primary investigator and researchers named in this study. If the results of this project are published your identity will not be made public. All laboratory notebooks and laboratory reports will be kept in a secure manner for six years as required by Oregon State University System Administration Rules. (http://www.ous.edu/archives/schdsection/grants.html#Labratory%20Notebooks).

DO I HAVE A CHOICE TO BE IN THE STUDY?

If you decide to take part in the study, it should be because you really want to volunteer. You will not lose any benefits or rights you would normally have if you choose not to volunteer. You can stop at any time during the study and still keep the benefits and rights you had before volunteering.

You will not be treated differently if you decide to stop taking part in the study. If you decide to not answer specific questions related to your health history we may decide to not include you in the study because we want to assure that subjects are healthy enough to participate in this study for their own safety. If you choose to withdraw from this project before it ends, the researchers may keep information collected about you and this information may be included in study reports.

WHAT IF I HAVE QUESTIONS?

If you have any questions about this research project, please contact: Mark A. Hoffman, 541-737-6787, mark.hoffman@oregonstate.edu, or Grace M. Golden, 541-231-2182, goldeng@onid.orst.edu.

If you have questions about your rights as a participant, please contact the Oregon State University Institutional Review Board (IRB) Human Protections Administrator, at (541) 737-4933 or by email at IRB@oregonstate.edu.

Your signature indicates that this research study has been explained to you, that your questions have been answered, and that you agree to take part in this study. You will receive a copy of this form.

Participant's Name (printed): ____

____ ______________________________

(Signature of Participant) (Date)

APPENDIX D

COOPERATING INSTITUTION
LETTER OF APPROVAL

Project Title:	Knee and Hip Kinetics and Kinematics as the Result of Combining a Lateral False Step with a Sidestep Cutting Maneuver
Principle Investigator:	Mark A. Hoffman
Research Staff:	Grace M. Golden, Mike Pavol

I have received information related to this research study. As a representative of the cooperating institution, my signature indicates my approval for recruitment and voluntary participation of subjects.

Cooperating Institution:

Position at Institution:

Signature: ______________________________ Date: ______________

APPENDIX E

SPSS OUTPUT Manuscript1

General Linear Model: KNEE ANGLE_ABDUCTION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	ka_abd_run
2	ka_abd_lfs
3	ka_abd_ssc
4	ka_abd_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
ka_abd_run	f	4.0841	2.42504	11
	m	3.1586	2.32426	10
	Total	3.6434	2.36553	21
ka_abd_lfs	f	6.2275	2.91441	11
	m	4.9169	1.91533	10
	Total	5.6034	2.51944	21
ka_abd_ssc	f	6.6674	2.47965	11
	m	5.5731	2.04269	10
	Total	6.1463	2.29470	21
ka_abd_ls	f	7.3614	2.74517	11
	m	6.7690	2.02977	10
	Total	7.0793	2.39037	21

Box's Test of Equality of Covariance Matrices(a)

Box's M	6.699
F	.514
df1	10
df2	1683.537
Sig.	.881

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.
a Design: Intercept+F_M
Within Subjects Design: cond

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.834	28.379(b)	3.000	17.000	.000	.834	85.137	1.000
	Wilks' Lambda	.166	28.379(b)	3.000	17.000	.000	.834	85.137	1.000
	Hotelling's Trace	5.008	28.379(b)	3.000	17.000	.000	.834	85.137	1.000
	Roy's Largest Root	5.008	28.379(b)	3.000	17.000	.000	.834	85.137	1.000
cond * F_M	Pillai's Trace	.060	.359(b)	3.000	17.000	.783	.060	1.077	.106
	Wilks' Lambda	.940	.359(b)	3.000	17.000	.783	.060	1.077	.106
	Hotelling's Trace	.063	.359(b)	3.000	17.000	.783	.060	1.077	.106
	Roy's Largest Root	.063	.359(b)	3.000	17.000	.783	.060	1.077	.106

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.566	10.097	5	.073	.783	.947	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	132.696	3	44.232	38.160	.000	.668	114.481	1.000
	Greenhouse-Geisser	132.696	2.348	56.526	38.160	.000	.668	89.582	1.000
	Huynh-Feldt	132.696	2.840	46.720	38.160	.000	.668	108.386	1.000
	Lower-bound	132.696	1.000	132.696	38.160	.000	.668	38.160	1.000
cond * F_M	Sphericity Assumed	1.444	3	.481	.415	.743	.021	1.245	.128
	Greenhouse-Geisser	1.444	2.348	.615	.415	.695	.021	.975	.118
	Huynh-Feldt	1.444	2.840	.508	.415	.732	.021	1.179	.125
	Lower-bound	1.444	1.000	1.444	.415	.527	.021	.415	.094
Error(cond)	Sphericity Assumed	66.069	57	1.159					
	Greenhouse-Geisser	66.069	44.603	1.481					
	Huynh-Feldt	66.069	53.965	1.224					
	Lower-bound	66.069	19.000	3.477					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	124.002	1	124.002	78.163	.000	.804	78.163	1.000
	Quadratic	5.300	1	5.300	4.501	.047	.192	4.501	.522
	Cubic	3.394	1	3.394	4.758	.042	.200	4.758	.544
cond * F_M	Linear	.387	1	.387	.244	.627	.013	.244	.076
	Quadratic	1.030	1	1.030	.875	.361	.044	.875	.144
	Cubic	.026	1	.026	.037	.850	.002	.037	.054
Error(cond)	Linear	30.143	19	1.586					
	Quadratic	22.374	19	1.178					
	Cubic	13.552	19	.713					

a Computed using alpha = .05

Levene's Test of Equality of Error Variances(a)

	F	df1	df2	Sig.
ka_abd_run	.000	1	19	.995
ka_abd_lfs	1.820	1	19	.193
ka_abd_ssc	.708	1	19	.411
ka_abd_ls	.151	1	19	.702

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.
a Design: Intercept+F_M
Within Subjects Design: cond

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	2623.331	1	2623.331	134.424	.000	.876	134.424	1.000
F_M	20.152	1	20.152	1.033	.322	.052	1.033	.162
Error	370.792	19	19.515					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	6.085	.666	4.691	7.479
m	5.104	.698	3.642	6.566

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	3.621	.519	2.534	4.709
2	5.572	.544	4.433	6.712
3	6.120	.499	5.076	7.164
4	7.065	.531	5.953	8.178

3. F_M * cond

Measure: MEASURE_1

F_M	cond	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
f	1	4.084	.717	2.584	5.585
	2	6.228	.751	4.655	7.800
	3	6.667	.688	5.227	8.108
	4	7.361	.733	5.826	8.897
m	1	3.159	.752	1.585	4.732
	2	4.917	.788	3.268	6.566
	3	5.573	.722	4.062	7.084
	4	6.769	.769	5.159	8.379

General Linear Model: KNEE ANGLE_INTERNAL ROTATION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	ka_ir_run
2	ka_ir_lfs
3	ka_ir_ssc
4	ka_ir_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
ka_ir_run	f	8.3108	5.23932	11
	m	6.5731	6.22758	10
	Total	7.4833	5.65405	21
ka_ir_lfs	f	10.3839	5.96144	11
	m	6.3267	6.78390	10
	Total	8.4519	6.54143	21
ka_ir_ssc	f	10.1961	8.04711	11
	m	7.3751	5.88999	10
	Total	8.8527	7.07628	21
ka_ir_ls	f	10.8926	7.82012	11
	m	7.9805	5.65749	10
	Total	9.5059	6.87032	21

Box's Test of Equality of Covariance Matrices(a)

Box's M	8.700
F	.667
df1	10
df2	1683.537
Sig.	.756

Tests the null hypothesis that the observed covariance matrices of the dependent variables are equal across groups.
a Design: Intercept+F_M
Within Subjects Design: cond

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.105	.667(b)	3.000	17.000	.584	.105	2.001	.161
	Wilks' Lambda	.895	.667(b)	3.000	17.000	.584	.105	2.001	.161
	Hotelling's Trace	.118	.667(b)	3.000	17.000	.584	.105	2.001	.161
	Roy's Largest Root	.118	.667(b)	3.000	17.000	.584	.105	2.001	.161
cond * F_M	Pillai's Trace	.130	.847(b)	3.000	17.000	.487	.130	2.542	.195
	Wilks' Lambda	.870	.847(b)	3.000	17.000	.487	.130	2.542	.195
	Hotelling's Trace	.150	.847(b)	3.000	17.000	.487	.130	2.542	.195
	Roy's Largest Root	.150	.847(b)	3.000	17.000	.487	.130	2.542	.195

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.487	12.751	5	.026	.676	.798	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	43.980	3	14.660	1.391	.255	.068	4.172	.350
	Greenhouse-Geisser	43.980	2.029	21.675	1.391	.261	.068	2.822	.282
	Huynh-Feldt	43.980	2.393	18.379	1.391	.260	.068	3.328	.309
	Lower-bound	43.980	1.000	43.980	1.391	.253	.068	1.391	.202
cond * F_M	Sphericity Assumed	14.118	3	4.706	.446	.721	.023	1.339	.134
	Greenhouse-Geisser	14.118	2.029	6.958	.446	.646	.023	.906	.118
	Huynh-Feldt	14.118	2.393	5.900	.446	.678	.023	1.068	.124
	Lower-bound	14.118	1.000	14.118	.446	.512	.023	.446	.097
Error(cond)	Sphericity Assumed	600.888	57	10.542					
	Greenhouse-Geisser	600.888	38.552	15.586					
	Huynh-Feldt	600.888	45.466	13.216					
	Lower-bound	600.888	19.000	31.626					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	43.100	1	43.100	2.088	.165	.099	2.088	.279
	Quadratic	.360	1	.360	.087	.771	.005	.087	.059
	Cubic	.519	1	.519	.076	.786	.004	.076	.058
cond * F_M	Linear	1.370	1	1.370	.066	.800	.003	.066	.057
	Quadratic	6.504	1	6.504	1.572	.225	.076	1.572	.222
	Cubic	6.245	1	6.245	.913	.351	.046	.913	.148
Error(cond)	Linear	392.274	19	20.646					
	Quadratic	78.596	19	4.137					
	Cubic	130.019	19	6.843					

a Computed using alpha = .05

Levene's Test of Equality of Error Variances(a)

	F	df1	df2	Sig.
ka_ir_run	.806	1	19	.381
ka_ir_lfs	.394	1	19	.538
ka_ir_ssc	.514	1	19	.482
ka_ir_ls	.641	1	19	.433

Tests the null hypothesis that the error variance of the dependent variable is equal across groups.
a Design: Intercept+F_M
Within Subjects Design: cond

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	6062.149	1	6062.149	43.438	.000	.696	43.438	1.000
F_M	174.032	1	174.032	1.247	.278	.062	1.247	.186
Error	2651.636	19	139.560					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	9.946	1.781	6.218	13.673
m	7.064	1.868	3.154	10.973

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	7.442	1.252	4.822	10.061
2	8.355	1.390	5.445	11.265
3	8.786	1.553	5.536	12.035
4	9.437	1.503	6.290	12.583

3. F_M * cond

Measure: MEASURE_1

F_M	cond	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
f	1	8.311	1.727	4.696	11.926
	2	10.384	1.919	6.368	14.400
	3	10.196	2.143	5.711	14.681
	4	10.893	2.075	6.550	15.235
m	1	6.573	1.812	2.781	10.365
	2	6.327	2.013	2.114	10.539
	3	7.375	2.248	2.671	12.079
	4	7.981	2.176	3.426	12.535

General Linear Model: HIP ADDUCTION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	ha_add_run
2	ha_add_lfs
3	ha_add_ssc
4	ha_add_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
ha_add_run	f	-9.8819	2.43366	11
	m	-4.6274	2.26264	10
	Total	-7.3797	3.53502	21
ha_add_lfs	f	5.7689	3.13171	11
	m	9.9470	3.79022	10
	Total	7.7585	3.99253	21
ha_add_ssc	f	4.3366	3.75707	11
	m	6.9922	4.36190	10
	Total	5.6012	4.17931	21
ha_add_ls	f	14.7673	3.10011	11
	m	18.4812	4.16903	10
	Total	16.5358	4.02979	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.981	291.601(b)	3.000	17.000	.000	.981	874.804	1.000
	Wilks' Lambda	.019	291.601(b)	3.000	17.000	.000	.981	874.804	1.000
	Hotelling's Trace	51.459	291.601(b)	3.000	17.000	.000	.981	874.804	1.000
	Roy's Largest Root	51.459	291.601(b)	3.000	17.000	.000	.981	874.804	1.000
cond * F_M	Pillai's Trace	.130	.847(b)	3.000	17.000	.487	.130	2.540	.195
	Wilks' Lambda	.870	.847(b)	3.000	17.000	.487	.130	2.540	.195
	Hotelling's Trace	.149	.847(b)	3.000	17.000	.487	.130	2.540	.195
	Roy's Largest Root	.149	.847(b)	3.000	17.000	.487	.130	2.540	.195

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.854	2.794	5	.732	.919	1.000	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	6114.270	3	2038.090	363.845	.000	.950	1091.535	1.000
	Greenhouse-Geisser	6114.270	2.757	2217.617	363.845	.000	.950	1003.170	1.000
	Huynh-Feldt	6114.270	3.000	2038.090	363.845	.000	.950	1091.535	1.000
	Lower-bound	6114.270	1.000	6114.270	363.845	.000	.950	363.845	1.000
cond * F_M	Sphericity Assumed	18.254	3	6.085	1.086	.362	.054	3.259	.278
	Greenhouse-Geisser	18.254	2.757	6.621	1.086	.360	.054	2.995	.266
	Huynh-Feldt	18.254	3.000	6.085	1.086	.362	.054	3.259	.278
	Lower-bound	18.254	1.000	18.254	1.086	.310	.054	1.086	.168
Error(cond)	Sphericity Assumed	319.287	57	5.602					
	Greenhouse-Geisser	319.287	52.386	6.095					
	Huynh-Feldt	319.287	57.000	5.602					
	Lower-bound	319.287	19.000	16.805					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	5051.978	1	5051.978	872.144	.000	.979	872.144	1.000
	Quadratic	90.330	1	90.330	14.802	.001	.438	14.802	.954
	Cubic	971.962	1	971.962	197.984	.000	.912	197.984	1.000
cond * F_M	Linear	9.887	1	9.887	1.707	.207	.082	1.707	.237
	Quadratic	5.967	1	5.967	.978	.335	.049	.978	.156
	Cubic	2.399	1	2.399	.489	.493	.025	.489	.102
Error(cond)	Linear	110.059	19	5.793					
	Quadratic	115.951	19	6.103					
	Cubic	93.277	19	4.909					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	2744.984	1	2744.984	90.185	.000	.826	90.185	1.000
F_M	327.002	1	327.002	10.743	.004	.361	10.743	.875
Error	578.308	19	30.437					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	3.748	.832	2.007	5.489
m	7.698	.872	5.872	9.524

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-7.255	.514	-8.331	-6.178
2	7.858	.756	6.276	9.440
3	5.664	.886	3.810	7.518
4	16.624	.796	14.957	18.291

General Linear Model: Hip angle_INTERNAL ROTATION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	ha_ir_run
2	ha_ir_lfs
3	ha_ir_ssc
4	ha_ir_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
ha_ir_run	f	-2.2685	5.98620	11
	m	-6.0324	3.64998	10
	Total	-4.0609	5.25573	21
ha_ir_lfs	f	-2.2540	8.96454	11
	m	-3.9897	5.43869	10
	Total	-3.0805	7.36758	21
ha_ir_ssc	f	.9805	7.05010	11
	m	-.1519	7.14956	10
	Total	.4413	6.94191	21
ha_ir_ls	f	1.4409	9.10934	11
	m	.6248	8.43743	10
	Total	1.0523	8.58487	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.434	4.350(b)	3.000	17.000	.019	.434	13.051	.777
	Wilks' Lambda	.566	4.350(b)	3.000	17.000	.019	.434	13.051	.777
	Hotelling's Trace	.768	4.350(b)	3.000	17.000	.019	.434	13.051	.777
	Roy's Largest Root	.768	4.350(b)	3.000	17.000	.019	.434	13.051	.777
cond * F_M	Pillai's Trace	.063	.381(b)	3.000	17.000	.768	.063	1.142	.110
	Wilks' Lambda	.937	.381(b)	3.000	17.000	.768	.063	1.142	.110
	Hotelling's Trace	.067	.381(b)	3.000	17.000	.768	.063	1.142	.110
	Roy's Largest Root	.067	.381(b)	3.000	17.000	.768	.063	1.142	.110

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.569	9.983	5	.076	.760	.914	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	413.346	3	137.782	7.651	.000	.287	22.952	.983
	Greenhouse-Geisser	413.346	2.279	181.390	7.651	.001	.287	17.434	.952
	Huynh-Feldt	413.346	2.742	150.731	7.651	.000	.287	20.980	.975
	Lower-bound	413.346	1.000	413.346	7.651	.012	.287	7.651	.747
cond * F_M	Sphericity Assumed	27.549	3	9.183	.510	.677	.026	1.530	.148
	Greenhouse-Geisser	27.549	2.279	12.089	.510	.628	.026	1.162	.133
	Huynh-Feldt	27.549	2.742	10.046	.510	.661	.026	1.398	.143
	Lower-bound	27.549	1.000	27.549	.510	.484	.026	.510	.104
Error(cond)	Sphericity Assumed	1026.522	57	18.009					
	Greenhouse-Geisser	1026.522	43.297	23.709					
	Huynh-Feldt	1026.522	52.103	19.702					
	Lower-bound	1026.522	19.000	54.027					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	381.633	1	381.633	12.273	.002	.392	12.273	.913
	Quadratic	.881	1	.881	.134	.718	.007	.134	.064
	Cubic	30.832	1	30.832	1.885	.186	.090	1.885	.257
cond * F_M	Linear	23.372	1	23.372	.752	.397	.038	.752	.131
	Quadratic	3.837	1	3.837	.583	.454	.030	.583	.112
	Cubic	.339	1	.339	.021	.887	.001	.021	.052
Error(cond)	Linear	590.827	19	31.096					
	Quadratic	125.000	19	6.579					
	Cubic	310.695	19	16.352					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	177.739	1	177.739	1.145	.298	.057	1.145	.174
F_M	72.647	1	72.647	.468	.502	.024	.468	.100
Error	2949.165	19	155.219					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	-.525	1.878	-4.456	3.406
m	-2.387	1.970	-6.510	1.736

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-4.150	1.096	-6.445	-1.856
2	-3.122	1.639	-6.553	.309
3	.414	1.551	-2.831	3.660
4	1.033	1.922	-2.990	5.056

General Linear Model: KNEE MOMENT_EXTENSION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	km_ext_run
2	km_ext_lfs
3	km_ext_ssc
4	km_ext_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
km_ext_run	f	-5.1596	2.65708	11
	m	-5.6472	2.14418	10
	Total	-5.3918	2.37933	21
km_ext_lfs	f	-4.4617	2.34531	11
	m	-4.9606	2.90714	10
	Total	-4.6993	2.57266	21
km_ext_ssc	f	-5.2341	4.18817	11
	m	-6.0469	4.40706	10
	Total	-5.6211	4.20516	21
km_ext_ls	f	-4.8184	3.17989	11
	m	-5.2241	3.02500	10
	Total	-5.0116	3.03591	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.178	1.229(b)	3.000	17.000	.330	.178	3.688	.271
	Wilks' Lambda	.822	1.229(b)	3.000	17.000	.330	.178	3.688	.271
	Hotelling's Trace	.217	1.229(b)	3.000	17.000	.330	.178	3.688	.271
	Roy's Largest Root	.217	1.229(b)	3.000	17.000	.330	.178	3.688	.271
cond * F_M	Pillai's Trace	.006	.037(b)	3.000	17.000	.990	.006	.110	.055
	Wilks' Lambda	.994	.037(b)	3.000	17.000	.990	.006	.110	.055
	Hotelling's Trace	.006	.037(b)	3.000	17.000	.990	.006	.110	.055
	Roy's Largest Root	.006	.037(b)	3.000	17.000	.990	.006	.110	.055

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.624	8.362	5	.138	.770	.928	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	10.605	3	3.535	.845	.475	.043	2.534	.222
	Greenhouse-Geisser	10.605	2.309	4.593	.845	.451	.043	1.950	.196
	Huynh-Feldt	10.605	2.785	3.808	.845	.468	.043	2.353	.214
	Lower-bound	10.605	1.000	10.605	.845	.370	.043	.845	.141
cond * F_M	Sphericity Assumed	.505	3	.168	.040	.989	.002	.121	.057
	Greenhouse-Geisser	.505	2.309	.219	.040	.974	.002	.093	.056
	Huynh-Feldt	.505	2.785	.181	.040	.986	.002	.112	.056
	Lower-bound	.505	1.000	.505	.040	.843	.002	.040	.054
Error(cond)	Sphericity Assumed	238.508	57	4.184					
	Greenhouse-Geisser	238.508	43.866	5.437					
	Huynh-Feldt	238.508	52.912	4.508					
	Lower-bound	238.508	19.000	12.553					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	.049	1	.049	.012	.912	.001	.012	.051
	Quadratic	.028	1	.028	.006	.938	.000	.006	.051
	Cubic	10.528	1	10.528	2.570	.125	.119	2.570	.331
cond * F_M	Linear	.001	1	.001	.000	.986	.000	.000	.050
	Quadratic	.229	1	.229	.051	.824	.003	.051	.055
	Cubic	.275	1	.275	.067	.799	.004	.067	.057
Error(cond)	Linear	75.073	19	3.951					
	Quadratic	85.592	19	4.505					
	Cubic	77.844	19	4.097					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	2261.053	1	2261.053	79.819	.000	.808	79.819	1.000
F_M	6.367	1	6.367	.225	.641	.012	.225	.074
Error	538.218	19	28.327					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	-4.918	.802	-6.598	-3.239
m	-5.470	.842	-7.231	-3.708

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-5.403	.530	-6.513	-4.293
2	-4.711	.574	-5.912	-3.510
3	-5.640	.938	-7.604	-3.677
4	-5.021	.679	-6.442	-3.600

General Linear Model: KNEE MOMENT_ADDUCTION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	km_add_run
2	km_add_lfs
3	km_add_ssc
4	km_add_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
km_add_run	f	.2258	.39933	11
	m	.0347	.65214	10
	Total	.1348	.52979	21
km_add_lfs	f	-1.2054	1.01732	11
	m	-1.1058	.57446	10
	Total	-1.1579	.81766	21
km_add_ssc	f	-1.8070	1.13073	11
	m	-2.2647	.88969	10
	Total	-2.0249	1.02486	21
km_add_ls	f	-3.6187	1.89998	11
	m	-4.2478	1.51565	10
	Total	-3.9182	1.71533	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.895	48.554(b)	3.000	17.000	.000	.895	145.662	1.000
	Wilks' Lambda	.105	48.554(b)	3.000	17.000	.000	.895	145.662	1.000
	Hotelling's Trace	8.568	48.554(b)	3.000	17.000	.000	.895	145.662	1.000
	Roy's Largest Root	8.568	48.554(b)	3.000	17.000	.000	.895	145.662	1.000
cond * F_M	Pillai's Trace	.109	.693(b)	3.000	17.000	.569	.109	2.080	.166
	Wilks' Lambda	.891	.693(b)	3.000	17.000	.569	.109	2.080	.166
	Hotelling's Trace	.122	.693(b)	3.000	17.000	.569	.109	2.080	.166
	Roy's Largest Root	.122	.693(b)	3.000	17.000	.569	.109	2.080	.166

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.567	10.045	5	.074	.734	.878	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	183.059	3	61.020	76.426	.000	.801	229.278	1.000
	Greenhouse-Geisser	183.059	2.202	83.142	76.426	.000	.801	168.271	1.000
	Huynh-Feldt	183.059	2.633	69.514	76.426	.000	.801	201.262	1.000
	Lower-bound	183.059	1.000	183.059	76.426	.000	.801	76.426	1.000
cond * F_M	Sphericity Assumed	1.595	3	.532	.666	.576	.034	1.998	.182
	Greenhouse-Geisser	1.595	2.202	.725	.666	.533	.034	1.466	.159
	Huynh-Feldt	1.595	2.633	.606	.666	.558	.034	1.754	.172
	Lower-bound	1.595	1.000	1.595	.666	.425	.034	.666	.121
Error(cond)	Sphericity Assumed	45.510	57	.798					
	Greenhouse-Geisser	45.510	41.833	1.088					
	Huynh-Feldt	45.510	50.035	.910					
	Lower-bound	45.510	19.000	2.395					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	178.979	1	178.979	145.637	.000	.885	145.637	1.000
	Quadratic	1.959	1	1.959	2.614	.122	.121	2.614	.336
	Cubic	2.120	1	2.120	5.086	.036	.211	5.086	.572
cond * F_M	Linear	.917	1	.917	.746	.398	.038	.746	.130
	Quadratic	.280	1	.280	.373	.548	.019	.373	.089
	Cubic	.399	1	.399	.956	.340	.048	.956	.153
Error(cond)	Linear	23.350	19	1.229					
	Quadratic	14.238	19	.749					
	Cubic	7.921	19	.417					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	256.254	1	256.254	97.541	.000	.837	97.541	1.000
F_M	1.818	1	1.818	.692	.416	.035	.692	.124
Error	49.915	19	2.627					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	-1.601	.244	-2.113	-1.090
m	-1.896	.256	-2.432	-1.359

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.130	.117	-.114	.375
2	-1.156	.183	-1.538	-.773
3	-2.036	.224	-2.504	-1.568
4	-3.933	.378	-4.724	-3.143

General Linear Model: KNEE MOMENT EXTERNAL ROTATION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	km_er_run
2	km_er_lfs
3	km_er_ssc
4	km_er_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
km_er_run	f	-.1321	.14737	11
	m	-.1393	.13259	10
	Total	-.1355	.13705	21
km_er_lfs	f	-.2264	.15838	11
	m	-.1970	.14843	10
	Total	-.2124	.15061	21
km_er_ssc	f	-.1570	.16896	11
	m	-.0514	.15898	10
	Total	-.1068	.16902	21
km_er_ls	f	-.2460	.30551	11
	m	-.0894	.28665	10
	Total	-.1714	.30011	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.382	3.508(b)	3.000	17.000	.038	.382	10.525	.676
	Wilks' Lambda	.618	3.508(b)	3.000	17.000	.038	.382	10.525	.676
	Hotelling's Trace	.619	3.508(b)	3.000	17.000	.038	.382	10.525	.676
	Roy's Largest Root	.619	3.508(b)	3.000	17.000	.038	.382	10.525	.676
cond * F_M	Pillai's Trace	.077	.474(b)	3.000	17.000	.705	.077	1.421	.126
	Wilks' Lambda	.923	.474(b)	3.000	17.000	.705	.077	1.421	.126
	Hotelling's Trace	.084	.474(b)	3.000	17.000	.705	.077	1.421	.126
	Roy's Largest Root	.084	.474(b)	3.000	17.000	.705	.077	1.421	.126

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.387	16.824	5	.005	.662	.778	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	.132	3	.044	1.435	.242	.070	4.305	.360
	Greenhouse-Geisser	.132	1.987	.067	1.435	.251	.070	2.851	.287
	Huynh-Feldt	.132	2.335	.057	1.435	.248	.070	3.351	.313
	Lower-bound	.132	1.000	.132	1.435	.246	.070	1.435	.207
cond * F_M	Sphericity Assumed	.086	3	.029	.929	.433	.047	2.786	.242
	Greenhouse-Geisser	.086	1.987	.043	.929	.403	.047	1.845	.198
	Huynh-Feldt	.086	2.335	.037	.929	.415	.047	2.168	.213
	Lower-bound	.086	1.000	.086	.929	.347	.047	.929	.150
Error(cond)	Sphericity Assumed	1.754	57	.031					
	Greenhouse-Geisser	1.754	37.752	.046					
	Huynh-Feldt	1.754	44.365	.040					
	Lower-bound	1.754	19.000	.092					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	.000	1	.000	.002	.962	.000	.002	.050
	Quadratic	.001	1	.001	.048	.829	.003	.048	.055
	Cubic	.132	1	.132	7.717	.012	.289	7.717	.750
cond * F_M	Linear	.084	1	.084	1.450	.243	.071	1.450	.208
	Quadratic	.000	1	.000	.016	.902	.001	.016	.052
	Cubic	.001	1	.001	.064	.802	.003	.064	.057
Error(cond)	Linear	1.105	19	.058					
	Quadratic	.325	19	.017					
	Cubic	.324	19	.017					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	2.009	1	2.009	30.381	.000	.615	30.381	.999
F_M	.106	1	.106	1.601	.221	.078	1.601	.225
Error	1.257	19	.066					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	-.190	.039	-.272	-.109
m	-.119	.041	-.204	-.034

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-.136	.031	-.200	-.071
2	-.212	.034	-.282	-.141
3	-.104	.036	-.179	-.029
4	-.168	.065	-.303	-.032

General Linear Model: HIP MOMENT_ABDUCTION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	hm_abd_run
2	hm_abd_lfs
3	hm_abd_ssc
4	hm_abd_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
hm_abd_run	f	5.4040	2.17129	11
	m	5.5864	2.42074	10
	Total	5.4909	2.23673	21
hm_abd_lfs	f	5.8635	3.47663	11
	m	5.7468	3.81854	10
	Total	5.8079	3.55086	21
hm_abd_ssc	f	4.6539	3.25831	11
	m	3.9367	4.29251	10
	Total	4.3124	3.70602	21
hm_abd_ls	f	3.4977	3.18006	11
	m	1.9870	3.81931	10
	Total	2.7783	3.49548	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.686	12.397(b)	3.000	17.000	.000	.686	37.191	.998
	Wilks' Lambda	.314	12.397(b)	3.000	17.000	.000	.686	37.191	.998
	Hotelling's Trace	2.188	12.397(b)	3.000	17.000	.000	.686	37.191	.998
	Roy's Largest Root	2.188	12.397(b)	3.000	17.000	.000	.686	37.191	.998
cond * F_M	Pillai's Trace	.117	.748(b)	3.000	17.000	.538	.117	2.245	.176
	Wilks' Lambda	.883	.748(b)	3.000	17.000	.538	.117	2.245	.176
	Hotelling's Trace	.132	.748(b)	3.000	17.000	.538	.117	2.245	.176
	Roy's Largest Root	.132	.748(b)	3.000	17.000	.538	.117	2.245	.176

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.683	6.752	5	.240	.796	.966	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	121.452	3	40.484	9.549	.000	.334	28.646	.996
	Greenhouse-Geisser	121.452	2.387	50.882	9.549	.000	.334	22.792	.986
	Huynh-Feldt	121.452	2.897	41.926	9.549	.000	.334	27.661	.995
	Lower-bound	121.452	1.000	121.452	9.549	.006	.334	9.549	.834
cond * F_M	Sphericity Assumed	8.772	3	2.924	.690	.562	.035	2.069	.187
	Greenhouse-Geisser	8.772	2.387	3.675	.690	.531	.035	1.646	.169
	Huynh-Feldt	8.772	2.897	3.028	.690	.557	.035	1.998	.184
	Lower-bound	8.772	1.000	8.772	.690	.417	.035	.690	.124
Error(cond)	Sphericity Assumed	241.664	57	4.240					
	Greenhouse-Geisser	241.664	45.352	5.329					
	Huynh-Feldt	241.664	55.040	4.391					
	Lower-bound	241.664	19.000	12.719					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	99.968	1	99.968	20.194	.000	.515	20.194	.989
	Quadratic	18.178	1	18.178	3.424	.080	.153	3.424	.420
	Cubic	3.307	1	3.307	1.344	.261	.066	1.344	.196
cond * F_M	Linear	8.449	1	8.449	1.707	.207	.082	1.707	.237
	Quadratic	.320	1	.320	.060	.809	.003	.060	.056
	Cubic	.003	1	.003	.001	.972	.000	.001	.050
Error(cond)	Linear	94.059	19	4.950					
	Quadratic	100.864	19	5.309					
	Cubic	46.742	19	2.460					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	1761.477	1	1761.477	54.443	.000	.741	54.443	1.000
F_M	6.123	1	6.123	.189	.668	.010	.189	.070
Error	614.731	19	32.354					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	4.855	.858	3.060	6.650
m	4.314	.899	2.432	6.197

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	5.495	.501	4.447	6.544
2	5.805	.796	4.140	7.471
3	4.295	.827	2.565	6.025
4	2.742	.764	1.143	4.342

General Linear Model: HIP MOMENT_EXTERNAL ROTATION

Within-Subjects Factors

Measure: MEASURE_1

cond	Dependent Variable
1	hm_er_run
2	hm_er_lfs
3	hm_er_ssc
4	hm_er_ls

Between-Subjects Factors

		N
F_M	f	11
	m	10

Descriptive Statistics

	F_M	Mean	Std. Deviation	N
hm_er_run	f	.2515	.30292	11
	m	.0647	.25407	10
	Total	.1626	.28994	21
hm_er_lfs	f	.1018	.50395	11
	m	-.2069	.50471	10
	Total	-.0452	.51630	21
hm_er_ssc	f	-.3395	.49423	11
	m	-.7884	.47708	10
	Total	-.5533	.52662	21
hm_er_ls	f	-.9055	.71093	11
	m	-1.7203	.86067	10
	Total	-1.2935	.87174	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Pillai's Trace	.836	28.809(b)	3.000	17.000	.000	.836	86.426	1.000
	Wilks' Lambda	.164	28.809(b)	3.000	17.000	.000	.836	86.426	1.000
	Hotelling's Trace	5.084	28.809(b)	3.000	17.000	.000	.836	86.426	1.000
	Roy's Largest Root	5.084	28.809(b)	3.000	17.000	.000	.836	86.426	1.000
cond * F_M	Pillai's Trace	.172	1.179(b)	3.000	17.000	.347	.172	3.537	.260
	Wilks' Lambda	.828	1.179(b)	3.000	17.000	.347	.172	3.537	.260
	Hotelling's Trace	.208	1.179(b)	3.000	17.000	.347	.172	3.537	.260
	Roy's Largest Root	.208	1.179(b)	3.000	17.000	.347	.172	3.537	.260

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+F_M
Within Subjects Design: cond

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
cond	.633	8.101	5	.151	.773	.934	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+F_M
Within Subjects Design: cond

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Sphericity Assumed	26.926	3	8.975	52.561	.000	.734	157.682	1.000
	Greenhouse-Geisser	26.926	2.320	11.606	52.561	.000	.734	121.943	1.000
	Huynh-Feldt	26.926	2.801	9.613	52.561	.000	.734	147.223	1.000
	Lower-bound	26.926	1.000	26.926	52.561	.000	.734	52.561	1.000
cond * F_M	Sphericity Assumed	1.163	3	.388	2.269	.090	.107	6.808	.544
	Greenhouse-Geisser	1.163	2.320	.501	2.269	.108	.107	5.265	.471
	Huynh-Feldt	1.163	2.801	.415	2.269	.095	.107	6.357	.524
	Lower-bound	1.163	1.000	1.163	2.269	.148	.107	2.269	.299
Error(cond)	Sphericity Assumed	9.733	57	.171					
	Greenhouse-Geisser	9.733	44.081	.221					
	Huynh-Feldt	9.733	53.219	.183					
	Lower-bound	9.733	19.000	.512					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	cond	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
cond	Linear	25.404	1	25.404	89.869	.000	.825	89.869	1.000
	Quadratic	1.517	1	1.517	11.081	.004	.368	11.081	.884
	Cubic	.004	1	.004	.045	.833	.002	.045	.055
cond * F_M	Linear	1.073	1	1.073	3.797	.066	.167	3.797	.456
	Quadratic	.078	1	.078	.570	.460	.029	.570	.111
	Cubic	.011	1	.011	.121	.731	.006	.121	.063
Error(cond)	Linear	5.371	19	.283					
	Quadratic	2.602	19	.137					
	Cubic	1.760	19	.093					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	16.434	1	16.434	24.376	.000	.562	24.376	.997
F_M	4.052	1	4.052	6.011	.024	.240	6.011	.643
Error	12.809	19	.674					

a Computed using alpha = .05

Estimated Marginal Means

1. F_M

Measure: MEASURE_1

F_M	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
f	-.223	.124	-.482	.036
m	-.663	.130	-.934	-.391

2. cond

Measure: MEASURE_1

cond	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.158	.061	.030	.287
2	-.053	.110	-.283	.178
3	-.564	.106	-.786	-.342
4	-1.313	.172	-1.672	-.954

SPSS Output_Manuscript2

Regression: KA_ABD RUN, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_run	3.6434	2.36553	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		KneeABD_run	Sex2	SQA	SexSQA
Pearson Correlation	KneeABD_run	1.000	.200	.451	.369
	Sex2	.200	1.000	-.138	.857
	SQA	.451	-.138	1.000	.252
	SexSQA	.369	.857	.252	1.000
Sig. (1-tailed)	KneeABD_run	.	.192	.020	.050
	Sex2	.192	.	.276	.000
	SQA	.020	.276	.	.135
	SexSQA	.050	.000	.135	.
N	KneeABD_run	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_run

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.525(a)	.276	.148	2.18362	.276	2.157	3	17	.131

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	30.856	3	10.285	2.157	.131(a)
	Residual	81.059	17	4.768		
	Total	111.915	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: KneeABD_run

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	.472	1.989		.238	.815	-3.724	4.669
	Sex2	.664	2.627	.144	.253	.804	-4.880	6.207
	SQA	.282	.196	.436	1.440	.168	-.131	.695
	SexSQA	.063	.270	.136	.234	.818	-.506	.632

a Dependent Variable: KneeABD_run

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.073	-.038	-.659
		SQA	-.038	.038	.365
		Sex2	-.659	.365	6.903

a Dependent Variable: KneeABD_run

Regression: KA_ABD LFS, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_lfs	4.7661	3.02401	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		KneeABD_lfs	Sex2	SQA	SexSQA
Pearson Correlation	KneeABD_lfs	1.000	.519	.287	.589
	Sex2	.519	1.000	-.138	.857
	SQA	.287	-.138	1.000	.252
	SexSQA	.589	.857	.252	1.000
Sig. (1-tailed)	KneeABD_lfs	.	.008	.103	.002
	Sex2	.008	.	.276	.000
	SQA	.103	.276	.	.135
	SexSQA	.002	.000	.135	.
N	KneeABD_lfs	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_lfs

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.634(a)	.401	.296	2.53753	.401	3.801	3	17	.030

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	73.428	3	24.476	3.801	.030(a)
	Residual	109.464	17	6.439		
	Total	182.892	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: KneeABD_lfs

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	.472	2.311		.204	.840	-4.404	5.349
	Sex2	3.018	3.053	.511	.989	.337	-3.424	9.460
	SQA	.282	.227	.341	1.239	.232	-.198	.762
	SexSQA	.038	.313	.065	.123	.904	-.623	.700

a Dependent Variable: KneeABD_lfs

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.098	-.052	-.890
		SQA	-.052	.052	.493
		Sex2	-.890	.493	9.323

a Dependent Variable: KneeABD_lfs

Regression: KA_ABD SSC, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_ssc	6.1463	2.29470	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		KneeABD_ssc	Sex2	SQA	SexSQA
Pearson Correlation	KneeABD_ssc	1.000	.244	.465	.383
	Sex2	.244	1.000	-.138	.857
	SQA	.465	-.138	1.000	.252
	SexSQA	.383	.857	.252	1.000
Sig. (1-tailed)	KneeABD_ssc	.	.143	.017	.043
	Sex2	.143	.	.276	.000
	SQA	.017	.276	.	.135
	SexSQA	.043	.000	.135	.
N	KneeABD_ssc	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_ssc

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.561(a)	.314	.193	2.06107	.314	2.597	3	17	.086

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	33.097	3	11.032	2.597	.086(a)
	Residual	72.216	17	4.248		
	Total	105.313	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: KneeABD_ssc

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	2.277	1.877		1.213	.242	-1.684	6.238
	Sex2	1.883	2.480	.420	.759	.458	-3.350	7.115
	SQA	.346	.185	.552	1.872	.078	-.044	.736
	SexSQA	-.052	.255	-.116	-.206	.840	-.589	.485

a Dependent Variable: KneeABD_ssc

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.065	-.034	-.587
		SQA	-.034	.034	.325
		Sex2	-.587	.325	6.150

a Dependent Variable: KneeABD_ssc

Independent t-tests (Manuscript 2)

T-Test

Group Statistics

	MF	N	Mean	Std. Deviation	Std. Error Mean
Goniometer	1.00	11	9.969	3.8313	1.1552
	2.00	9	7.592	3.3119	1.1040

Independent Samples Test

		Levene's Test for Equality of Variances		t-test for Equality of Means						
		F	Sig.	t	df	Sig. (2-tailed)	Mean Difference	Std. Error Difference	95% Confidence Interval of the Difference	
									Lower	Upper
Goniometer	Equal variances assumed	.223	.643	1.465	18	.160	2.3769	1.6224	-1.0317	5.7854
	Equal variances not assumed			1.488	17.921	.154	2.3769	1.5979	-.9811	5.7349

T-Test

Group Statistics

	Sex	N	Mean	Std. Deviation	Std. Error Mean
Position	1.00	10	19.6482	1.51343	.47859
	2.00	10	19.4275	1.87189	.59194

Independent Samples Test

		Levene's Test for Equality of Variances		t-test for Equality of Means						
		F	Sig.	t	df	Sig. (2-tailed)	Mean Difference	Std. Error Difference	95% Confidence Interval of the Difference	
									Lower	Upper
Position	Equal variances assumed	.464	.504	.290	18	.775	.22063	.76121	-1.37861	1.81988
	Equal variances not assumed			.290	17.244	.775	.22063	.76121	-1.38366	1.82492

T-Test

Group Statistics

	Sex	N	Mean	Std. Deviation	Std. Error Mean
NorT_x_100	1.00	11	1.3164	.17964	.05416
	2.00	10	1.2487	.21579	.06824

Independent Samples Test

		Levene's Test for Equality of Variances		t-test for Equality of Means						
		F	Sig.	t	df	Sig. (2-tailed)	Mean Difference	Std. Error Difference	95% Confidence Interval of the Difference	
									Lower	Upper
NorT_x_100	Equal variances assumed	.113	.741	.785	19	.442	.06777	.08633	-.11293	.24847
	Equal variances not assumed			.778	17.619	.447	.06777	.08712	-.11555	.25109

Regression: KA_ABD LFS-SSC, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_ls	7.0793	2.39037	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		KneeABD_ls	Sex2	SQA	SexSQA
Pearson Correlation	KneeABD_ls	1.000	.127	.327	.206
	Sex2	.127	1.000	-.138	.857
	SQA	.327	-.138	1.000	.252
	SexSQA	.206	.857	.252	1.000
Sig. (1-tailed)	KneeABD_ls	.	.292	.074	.185
	Sex2	.292	.	.276	.000
	SQA	.074	.276	.	.135
	SexSQA	.185	.000	.135	.
N	KneeABD_ls	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_ls

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.381(a)	.145	-.006	2.39727	.145	.962	3	17	.433

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	16.580	3	5.527	.962	.433(a)
	Residual	97.697	17	5.747		
	Total	114.278	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: KneeABD_ls

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	3.983	2.184		1.824	.086	-.624	8.590
	Sex2	1.907	2.885	.408	.661	.517	-4.178	7.993
	SQA	.292	.215	.448	1.361	.191	-.161	.746
	SexSQA	-.120	.296	-.257	-.406	.690	-.745	.504

a Dependent Variable: KneeABD_ls

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.088	-.046	-.794
		SQA	-.046	.046	.440
		Sex2	-.794	.440	8.320

a Dependent Variable: KneeABD_ls

Regression: KM_ADD RUN, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_run	.1348	.52979	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		km_add_run	Sex2	SQA	SexSQA
Pearson Correlation	km_add_run	1.000	.185	-.223	.120
	Sex2	.185	1.000	-.138	.857
	SQA	-.223	-.138	1.000	.252
	SexSQA	.120	.857	.252	1.000
Sig. (1-tailed)	km_add_run	.	.211	.166	.302
	Sex2	.211	.	.276	.000
	SQA	.166	.276	.	.135
	SexSQA	.302	.000	.135	.
N	km_add_run	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_run

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.291(a)	.085	-.077	.54980	.085	.524	3	17	.672

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.475	3	.158	.524	.672(a)
	Residual	5.139	17	.302		
	Total	5.614	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: km_add_run

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	.464	.501		.927	.367	-.592	1.521
	Sex2	-.112	.662	-.108	-.169	.868	-1.508	1.284
	SQA	-.045	.049	-.312	-.915	.373	-.149	.059
	SexSQA	.030	.068	.292	.446	.661	-.113	.174

a Dependent Variable: km_add_run

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.005	-.002	-.042
		SQA	-.002	.002	.023
		Sex2	-.042	.023	.438

a Dependent Variable: km_add_run

Regression: KM_ADD LFS, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_lfs	-1.1579	.81766	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		km_add_lfs	Sex2	SQA	SexSQA
Pearson Correlation	km_add_lfs	1.000	-.062	.083	-.091
	Sex2	-.062	1.000	-.138	.857
	SQA	.083	-.138	1.000	.252
	SexSQA	-.091	.857	.252	1.000
Sig. (1-tailed)	km_add_lfs	.	.394	.361	.348
	Sex2	.394	.	.276	.000
	SQA	.361	.276	.	.135
	SexSQA	.348	.000	.135	.
N	km_add_lfs	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_lfs

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.208(a)	.043	-.125	.86743	.043	.257	3	17	.855

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.580	3	.193	.257	.855(a)
	Residual	12.791	17	.752		
	Total	13.371	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: km_add_lfs

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	-1.684	.790		-2.131	.048	-3.351	-.017
	Sex2	.671	1.044	.420	.642	.529	-1.532	2.873
	SQA	.061	.078	.272	.780	.446	-.103	.225
	SexSQA	-.083	.107	-.519	-.776	.448	-.309	.143

a Dependent Variable: km_add_lfs

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.011	-.006	-.104
		SQA	-.006	.006	.058
		Sex2	-.104	.058	1.089

a Dependent Variable: km_add_lfs

Regression: KM_ADD SSC, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_ssc	-2.0249	1.02486	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		km_add_ssc	Sex2	SQA	SexSQA
Pearson Correlation	km_add_ssc	1.000	.229	-.059	.242
	Sex2	.229	1.000	-.138	.857
	SQA	-.059	-.138	1.000	.252
	SexSQA	.242	.857	.252	1.000
Sig. (1-tailed)	km_add_ssc	.	.160	.400	.145
	Sex2	.160	.	.276	.000
	SQA	.400	.276	.	.135
	SexSQA	.145	.000	.135	.
N	km_add_ssc	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_ssc

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.281(a)	.079	-.084	1.06690	.079	.485	3	17	.697

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	1.656	3	.552	.485	.697(a)
	Residual	19.351	17	1.138		
	Total	21.007	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: km_add_ssc

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	-1.733	.972		-1.784	.092	-3.784	.317
	Sex2	-.375	1.284	-.187	-.292	.774	-3.083	2.334
	SQA	-.056	.096	-.199	-.583	.568	-.257	.146
	SexSQA	.091	.132	.453	.690	.499	-.187	.369

a Dependent Variable: km_add_ssc

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.017	-.009	-.157
		SQA	-.009	.009	.087
		Sex2	-.157	.087	1.648

a Dependent Variable: km_add_ssc

Regression: KM_ADD LFS-SSC, SQA

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_ls	-3.9182	1.71533	21
Sex2	.5238	.51177	21
SQA	9.0152	3.66145	21
SexSQA	4.4762	5.10399	21

Correlations

		km_add_ls	Sex2	SQA	SexSQA
Pearson Correlation	km_add_ls	1.000	.188	-.022	.308
	Sex2	.188	1.000	-.138	.857
	SQA	-.022	-.138	1.000	.252
	SexSQA	.308	.857	.252	1.000
Sig. (1-tailed)	km_add_ls	.	.208	.462	.088
	Sex2	.208	.	.276	.000
	SQA	.462	.276	.	.135
	SexSQA	.088	.000	.135	.
N	km_add_ls	21	21	21	21
	Sex2	21	21	21	21
	SQA	21	21	21	21
	SexSQA	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexSQA, SQA, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_ls

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.450(a)	.203	.062	1.66112	.203	1.442	3	17	.265

a Predictors: (Constant), SexSQA, SQA, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	11.938	3	3.979	1.442	.265(a)
	Residual	46.909	17	2.759		
	Total	58.847	20			

a Predictors: (Constant), SexSQA, SQA, Sex2
b Dependent Variable: km_add_ls

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	-2.318	1.513		-1.532	.144	-5.510	.874
	Sex2	-2.885	1.999	-.861	-1.443	.167	-7.102	1.332
	SQA	-.202	.149	-.432	-1.360	.192	-.516	.112
	SexSQA	.388	.205	1.154	1.891	.076	-.045	.821

a Dependent Variable: km_add_ls

Coefficient Correlations(a)

Model			SexSQA	SQA	Sex2
1	Correlations	SexSQA	1.000	-.726	-.930
		SQA	-.726	1.000	.710
		Sex2	-.930	.710	1.000
	Covariances	SexSQA	.042	-.022	-.381
		SQA	-.022	.022	.211
		Sex2	-.381	.211	3.995

a Dependent Variable: km_add_ls

Regression: KneeABD_RUN, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_run	3.6434	2.36553	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		KneeABD_run	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	KneeABD_run	1.000	.200	.225	.274
	Sex2	.200	1.000	.177	.983
	EccHipNorm	.225	.177	1.000	.294
	SexECCNorm	.274	.983	.294	1.000
Sig. (1-tailed)	KneeABD_run	.	.192	.163	.114
	Sex2	.192	.	.221	.000
	EccHipNorm	.163	.221	.	.098
	SexECCNorm	.114	.000	.098	.
N	KneeABD_run	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_run

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.477(a)	.227	.091	2.25513	.227	1.669	3	17	.211

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	25.459	3	8.486	1.669	.211(a)
	Residual	86.455	17	5.086		
	Total	111.915	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: KneeABD_run

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	5.420	4.408		1.230	.236	-3.880	14.720
	Sex2	-11.595	6.870	-2.508	-1.688	.110	-26.090	2.901
	EccHipNorm	-181.099	348.354	-.150	-.520	.610	-916.062	553.864
	SexECCNorm	960.395	528.151	2.783	1.818	.087	-153.906	2074.696

a Dependent Variable: KneeABD_run

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	278943.241	-121350.627	-3589.892
		EccHipNorm	-121350.627	121350.627	1515.273
		Sex2	-3589.892	1515.273	47.203

a Dependent Variable: KneeABD_run

Regression: KneeABD_LFS, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_lfs	4.7661	3.02401	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		KneeABD_lfs	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	KneeABD_lfs	1.000	.519	.304	.589
	Sex2	.519	1.000	.177	.983
	EccHipNorm	.304	.177	1.000	.294
	SexECCNorm	.589	.983	.294	1.000
Sig. (1-tailed)	KneeABD_lfs	.	.008	.090	.002
	Sex2	.008	.	.221	.000
	EccHipNorm	.090	.221	.	.098
	SexECCNorm	.002	.000	.098	.
N	KneeABD_lfs	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_lfs

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.677(a)	.458	.363	2.41421	.458	4.793	3	17	.013

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	83.809	3	27.936	4.793	.013(a)
	Residual	99.083	17	5.828		
	Total	182.892	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: KneeABD_lfs

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	5.420	4.719		1.149	.267	-4.536	15.376
	Sex2	-12.525	7.355	-2.120	-1.703	.107	-28.043	2.993
	EccHipNorm	-181.099	372.928	-.117	-.486	.633	-967.908	605.709
	SexECCNorm	1193.871	565.407	2.707	2.112	.050	.966	2386.776

a Dependent Variable: KneeABD_lfs

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	319685.378	-139074.964	-4114.228
		EccHipNorm	-139074.964	139074.964	1736.592
		Sex2	-4114.228	1736.592	54.097

a Dependent Variable: KneeABD_lfs

Regression: KneeABD_SSC, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_ssc	6.1463	2.29470	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		KneeABD_ssc	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	KneeABD_ssc	1.000	.244	.392	.325
	Sex2	.244	1.000	.177	.983
	EccHipNorm	.392	.177	1.000	.294
	SexECCNorm	.325	.983	.294	1.000
Sig. (1-tailed)	KneeABD_ssc	.	.143	.039	.075
	Sex2	.143	.	.221	.000
	EccHipNorm	.039	.221	.	.098
	SexECCNorm	.075	.000	.098	.
N	KneeABD_ssc	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_ssc

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.526(a)	.277	.149	2.11637	.277	2.171	3	17	.129

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	29.170	3	9.723	2.171	.129(a)
	Residual	76.143	17	4.479		
	Total	105.313	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: KneeABD_ssc

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	4.257	4.137		1.029	.318	-4.471	12.984
	Sex2	-8.565	6.448	-1.910	-1.328	.202	-22.168	5.038
	EccHipNorm	105.403	326.920	.090	.322	.751	-584.337	795.143
	SexECCNorm	728.319	495.653	2.176	1.469	.160	-317.418	1774.056

a Dependent Variable: KneeABD_ssc

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	245672.074	-106876.439	-3161.705
		EccHipNorm	-106876.439	106876.439	1334.537
		Sex2	-3161.705	1334.537	41.573

a Dependent Variable: KneeABD_ssc

Regression: KneeABD_LFS-SSC, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
KneeABD_ls	7.0793	2.39037	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		KneeABD_ls	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	KneeABD_ls	1.000	.127	.304	.190
	Sex2	.127	1.000	.177	.983
	EccHipNorm	.304	.177	1.000	.294
	SexECCNorm	.190	.983	.294	1.000
Sig. (1-tailed)	KneeABD_ls	.	.292	.090	.204
	Sex2	.292	.	.221	.000
	EccHipNorm	.090	.221	.	.098
	SexECCNorm	.204	.000	.098	.
N	KneeABD_ls	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: KneeABD_ls

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.382(a)	.146	-.004	2.39560	.146	.971	3	17	.429

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	16.716	3	5.572	.971	.429(a)
	Residual	97.561	17	5.739		
	Total	114.278	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: KneeABD_ls

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	5.335	4.682		1.139	.270	-4.544	15.214
	Sex2	-6.744	7.298	-1.444	-.924	.368	-22.142	8.654
	EccHipNorm	114.871	370.053	.094	.310	.760	-665.873	895.614
	SexECCNorm	551.393	561.049	1.581	.983	.339	-632.317	1735.102

a Dependent Variable: KneeABD_ls

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	314775.680	-136939.064	-4051.042
		EccHipNorm	-136939.064	136939.064	1709.921
		Sex2	-4051.042	1709.921	53.266

a Dependent Variable: KneeABD_ls

Regression: KM ADD_RUN, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_run	.1348	.52979	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		km_add_run	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	km_add_run	1.000	.185	.159	.239
	Sex2	.185	1.000	.177	.983
	EccHipNorm	.159	.177	1.000	.294
	SexECCNorm	.239	.983	.294	1.000
Sig. (1-tailed)	km_add_run	.	.211	.246	.149
	Sex2	.211	.	.221	.000
	EccHipNorm	.246	.221	.	.098
	SexECCNorm	.149	.000	.098	.
N	km_add_run	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_run

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.374(a)	.140	-.012	.53290	.140	.922	3	17	.451

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.786	3	.262	.922	.451(a)
	Residual	4.828	17	.284		
	Total	5.614	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: km_add_run

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	.497	1.042		.478	.639	-1.700	2.695
	Sex2	-1.970	1.624	-1.903	-1.213	.242	-5.395	1.456
	EccHipNorm	-37.063	82.319	-.137	-.450	.658	-210.740	136.615
	SexECCNorm	166.042	124.806	2.149	1.330	.201	-97.276	429.360

a Dependent Variable: km_add_run

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	15576.577	-6776.387	-200.465
		EccHipNorm	-6776.387	6776.387	84.615
		Sex2	-200.465	84.615	2.636

a Dependent Variable: km_add_run

Regression: KM ADD_LFS, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_lfs	-1.1579	.81766	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		km_add_lfs	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	km_add_lfs	1.000	-.062	.299	-.021
	Sex2	-.062	1.000	.177	.983
	EccHipNorm	.299	.177	1.000	.294
	SexECCNorm	-.021	.983	.294	1.000
Sig. (1-tailed)	km_add_lfs	.	.394	.094	.464
	Sex2	.394	.	.221	.000
	EccHipNorm	.094	.221	.	.098
	SexECCNorm	.464	.000	.098	.
N	km_add_lfs	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_lfs

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.322(a)	.103	-.055	.83974	.103	.654	3	17	.591

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	1.384	3	.461	.654	.591(a)
	Residual	11.988	17	.705		
	Total	13.371	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: km_add_lfs

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	-2.726	1.641		-1.661	.115	-6.188	.737
	Sex2	-.314	2.558	-.197	-.123	.904	-5.712	5.083
	EccHipNorm	129.716	129.716	.310	1.000	.331	-143.961	403.393
	SexECCNorm	9.627	196.667	.081	.049	.962	-405.303	424.558

a Dependent Variable: km_add_lfs

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	38677.746	-16826.250	-497.768
		EccHipNorm	-16826.250	16826.250	210.105
		Sex2	-497.768	210.105	6.545

a Dependent Variable: km_add_lfs

Regression: KM ADD_SSC, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_ssc	-2.0249	1.02486	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		km_add_ssc	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	km_add_ssc	1.000	.229	-.378	.158
	Sex2	.229	1.000	.177	.983
	EccHipNorm	-.378	.177	1.000	.294
	SexECCNorm	.158	.983	.294	1.000
Sig. (1-tailed)	km_add_ssc	.	.160	.045	.247
	Sex2	.160	.	.221	.000
	EccHipNorm	.045	.221	.	.098
	SexECCNorm	.247	.000	.098	.
N	km_add_ssc	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_ssc

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.494(a)	.244	.111	.96632	.244	1.832	3	17	.180

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	5.133	3	1.711	1.832	.180(a)
	Residual	15.874	17	.934		
	Total	21.007	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: km_add_ssc

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	-.046	1.889		-.025	.981	-4.031	3.938
	Sex2	2.056	2.944	1.027	.699	.494	-4.155	8.268
	EccHipNorm	-177.645	149.269	-.339	-1.190	.250	-492.574	137.285
	SexECCNorm	-112.302	226.311	-.751	-.496	.626	-589.777	365.172

a Dependent Variable: km_add_ssc

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	51216.622	-22281.125	-659.138
		EccHipNorm	-22281.125	22281.125	278.218
		Sex2	-659.138	278.218	8.667

a Dependent Variable: km_add_ssc

Regression: KM ADD_LFS, ECC

Descriptive Statistics

	Mean	Std. Deviation	N
km_add_ls	-3.9182	1.71533	21
Sex2	.5238	.51177	21
EccHipNorm	.012842	.0019568	21
SexECCNorm	.006896	.0068558	21

Correlations

		km_add_ls	Sex2	EccHipNorm	SexECCNorm
Pearson Correlation	km_add_ls	1.000	.188	-.189	.204
	Sex2	.188	1.000	.177	.983
	EccHipNorm	-.189	.177	1.000	.294
	SexECCNorm	.204	.983	.294	1.000
Sig. (1-tailed)	km_add_ls	.	.208	.205	.188
	Sex2	.208	.	.221	.000
	EccHipNorm	.205	.221	.	.098
	SexECCNorm	.188	.000	.098	.
N	km_add_ls	21	21	21	21
	Sex2	21	21	21	21
	EccHipNorm	21	21	21	21
	SexECCNorm	21	21	21	21

Variables Entered/Removed(b)

Model	Variables Entered	Variables Removed	Method
1	SexECCNorm, EccHipNorm, Sex2(a)	.	Enter

a All requested variables entered.
b Dependent Variable: km_add_ls

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
					R Square Change	F Change	df1	df2	Sig. F Change
1	.447(a)	.200	.059	1.66413	.200	1.416	3	17	.272

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2

ANOVA(b)

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	11.768	3	3.923	1.416	.272(a)
	Residual	47.079	17	2.769		
	Total	58.847	20			

a Predictors: (Constant), SexECCNorm, EccHipNorm, Sex2
b Dependent Variable: km_add_ls

Coefficients(a)

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95% Confidence Interval for B	
		B	Std. Error	Beta			Lower Bound	Upper Bound
1	(Constant)	1.557	3.253		.479	.638	-5.305	8.420
	Sex2	-7.026	5.070	-2.096	-1.386	.184	-17.723	3.671
	EccHipNorm	-464.899	257.062	-.530	-1.809	.088	-1007.252	77.454
	SexECCNorm	605.433	389.740	2.420	1.553	.139	-216.846	1427.711

a Dependent Variable: km_add_ls

Coefficient Correlations(a)

Model			SexECCNorm	EccHipNorm	Sex2
1	Correlations	SexECCNorm	1.000	-.660	-.989
		EccHipNorm	-.660	1.000	.633
		Sex2	-.989	.633	1.000
	Covariances	SexECCNorm	151896.905	-66080.772	-1954.855
		EccHipNorm	-66080.772	66080.772	825.133
		Sex2	-1954.855	825.133	25.704

a Dependent Variable: km_add_ls

Factor Analysis_Manuscript3

Correlation Matrix

		Knee EXT	Knee ABD	Knee IR	Hip FLX	Hip ABD	Hip ER	Foot Ang	MKne FLX	MKne EXT	MKne ABD	MKne ADD	MKne IR
Correl-ation	KneeEXT	1.00	.426	-.464	.233	-.068	-.382	.252	-.288	.735	-.502	.217	-.266
	KneeABD	.426	1.00	-.193	.317	.144	-.147	-.069	-.065	.371	-.578	-.409	.030
	KneeIR	-.464	-.193	1.00	-.301	.180	.530	-.477	.156	-.319	.341	-.195	.264
	HipFLX	.233	.317	-.301	1.00	-.212	-.138	.083	.286	.241	-.275	-.022	-.129
	HipABD	-.068	.144	.180	-.212	1.00	.138	-.143	.350	-.095	-.248	-.664	.353
	HipER	-.382	-.147	.530	-.138	.138	1.00	-.563	.118	-.137	-.029	-.318	.469
	FootAng	.252	-.069	-.477	.083	-.143	-.563	1.00	-.220	.108	.107	.361	-.371
	MKneFLX	-.288	-.065	.156	.286	.350	.118	-.220	1.00	-.196	.033	-.466	.218
	MKneEXT	.735	.371	-.319	.241	-.095	-.137	.108	-.196	1.00	-.532	.150	-.197
	MKneABD	-.502	-.578	.341	-.275	-.248	-.029	.107	.033	-.532	1.00	.341	-.071
	MKneADD	.217	-.409	-.195	-.022	-.664	-.318	.361	-.466	.150	.341	1.00	-.500
	MKneIR	-.266	.030	.264	-.129	.353	.469	-.371	.218	-.197	-.071	-.500	1.00
	MKneER	.441	.100	-.273	.295	-.124	.038	-.073	-.064	.384	-.499	.153	.205
	MHipEXT	.339	.150	.212	-.319	.086	.090	-.014	-.589	.388	-.274	.134	-.054
	MHipABD	-.620	-.377	.417	-.280	-.145	.164	-.028	.084	-.590	.871	.096	.171
	MHipADD	.264	-.319	-.223	-.045	-.621	-.282	.332	-.510	.161	.282	.958	-.482
	MHipIR	-.362	-.322	.274	-.124	-.304	-.167	.116	.036	-.474	.888	.291	.000
	MHipER	.228	-.286	.057	-.167	-.534	-.281	.219	-.509	.164	.345	.861	-.309

		MKne ER	MHip EXT	MHip ABD	MHip ADD	MHip IR	MHip ER
Correl-ation	KneeEXT	.441	.339	-.620	.264	-.362	.228
	KneeABD	.100	.150	-.377	-.319	-.322	-.286
	KneeIR	-.273	.212	.417	-.223	.274	.057
	HipFLX	.295	-.319	-.280	-.045	-.124	-.167
	HipABD	-.124	.086	-.145	-.621	-.304	-.534
	HipER	.038	.090	.164	-.282	-.167	-.281
	FootAng	-.073	-.014	-.028	.332	.116	.219
	MKneFLX	-.064	-.589	.084	-.510	.036	-.509
	MKneEXT	.384	.388	-.590	.161	-.474	.164
	MKneABD	-.499	-.274	.871	.282	.888	.345
	MKneADD	.153	.134	.096	.958	.291	.861
	MKneIR	.205	-.054	.171	-.482	.000	-.309
	MKneER	1.000	.179	-.552	.144	-.390	.071
	MHipEXT	.179	1.000	-.352	.109	-.314	.314
	MHipABD	-.552	-.352	1.000	.101	.847	.179
	MHipADD	.144	.109	.101	1.000	.243	.836
	MHipIR	-.390	-.314	.847	.243	1.000	.386
	MHipER	.071	.314	.179	.836	.386	1.000

Communalities

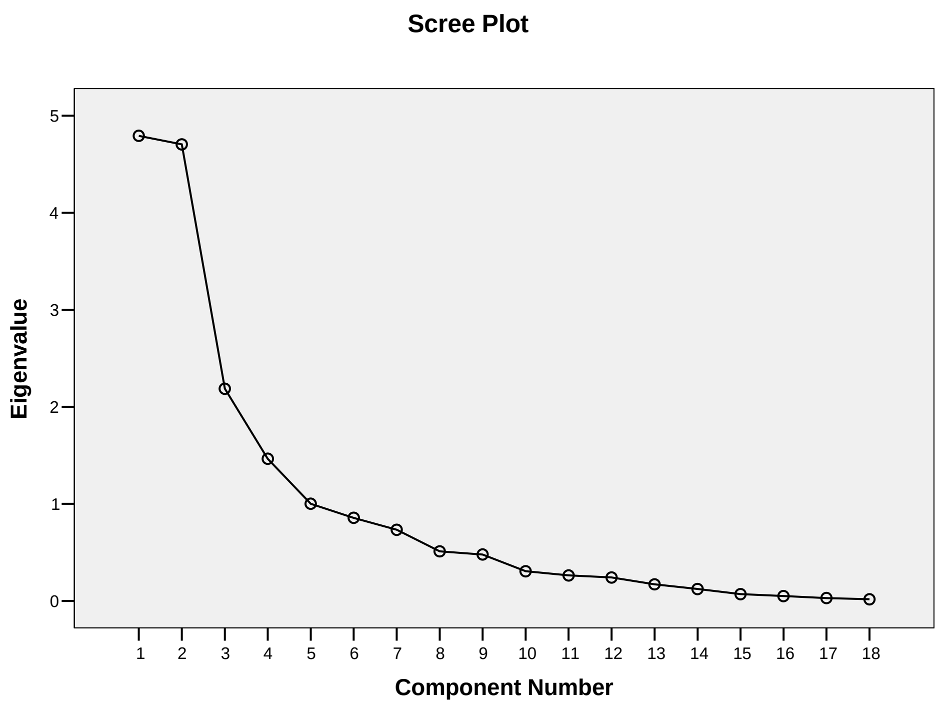
	Initial	Extraction
KneeEXT	1.000	.728
KneeABD	1.000	.456
KneeIR	1.000	.673
HipFLX	1.000	.682
HipABD	1.000	.668
HipER	1.000	.742
FootAng	1.000	.623
MKneFLX	1.000	.676
MKneEXT	1.000	.634
MKneABD	1.000	.904
MKneADD	1.000	.947
MKneIR	1.000	.507
MKneER	1.000	.702
MHipEXT	1.000	.843
MHipABD	1.000	.854
MHipADD	1.000	.896
MHipIR	1.000	.763
MHipER	1.000	.851

Extraction Method: Principal Component Analysis.

Total Variance Explained

Component	Initial Eigenvalues			Extraction Sums of Squared Loadings			Rotation Sums of Squared Loadings(a)
	Total	% of Variance	Cumulative %	Total	% of Variance	Cumulative %	Total
1	4.793	26.625	26.625	4.793	26.625	26.625	4.713
2	4.705	26.137	52.762	4.705	26.137	52.762	4.121
3	2.186	12.147	64.909	2.186	12.147	64.909	2.095
4	1.465	8.137	73.046	1.465	8.137	73.046	3.060
5	1.001	5.563	78.609				
6	.857	4.761	83.370				
7	.734	4.075	87.445				
8	.511	2.836	90.281				
9	.479	2.659	92.939				
10	.306	1.701	94.640				
11	.263	1.460	96.100				
12	.241	1.341	97.441				
13	.171	.951	98.392				
14	.123	.681	99.073				
15	.070	.389	99.462				
16	.050	.278	99.740				
17	.030	.165	99.905				
18	.017	.095	100.000				

Extraction Method: Principal Component Analysis.
a When components are correlated, sums of squared loadings cannot be added to obtain a total variance.



Component Matrix(a)

	Component			
	1	2	3	4
KneeEXT	-.813	.250	-.004	-.073
KneeABD	-.541	-.330	-.119	-.202
KneeIR	.563	-.242	.544	.054
HipFLX	-.357	-.066	-.563	.483
HipABD	-.006	-.687	.073	-.437
HipER	.274	-.467	.537	.401
FootAng	-.176	.493	-.424	-.411
MKneFLX	.251	-.552	-.488	.265
MKneEXT	-.773	.129	.131	.050
MKneABD	.847	.422	-.081	-.043
MKneADD	-.036	.947	.088	.204
MKneIR	.219	-.570	.263	.256
MKneER	-.583	.003	.149	.584
MHipEXT	-.382	.133	.763	-.313
MHipABD	.904	.185	-.048	-.023
MHipADD	-.070	.920	.104	.182
MHipIR	.745	.416	-.186	-.001
MHipER	.031	.863	.311	.086

Extraction Method: Principal Component Analysis.
a 4 components extracted.

Pattern Matrix(a)

	Component			
	1	2	3	4
KneeEXT	-.722	.161	.093	-.324
KneeABD	-.508	-.407	-.007	-.195
KneeIR	.373	-.043	.334	.598
HipFLX	-.319	.130	-.743	-.080
HipABD	-.053	-.801	.225	.022
HipER	.000	.000	.074	.859
FootAng	.075	.038	.021	-.783
MKneFLX	.207	-.381	-.631	.142
MKneEXT	-.754	.173	.097	-.097
MKneABD	.902	.278	.023	-.070
MKneADD	.064	.896	.088	-.191
MKneIR	.015	-.233	-.057	.615
MKneER	-.692	.410	-.253	.367
MHipEXT	-.449	.083	.806	.140
MHipABD	.907	.110	-.002	.071
MHipADD	.027	.866	.110	-.192
MHipIR	.818	.277	-.083	-.128
MHipER	.087	.805	.319	-.081

Extraction Method: Principal Component Analysis. , Rotation Method: Oblimin with Kaiser Normalization.
a Rotation converged in 11 iterations.

Structure Matrix

	Component			
	1	2	3	4
KneeEXT	-.750	.230	.086	-.439
KneeABD	-.538	-.373	-.065	-.156
KneeIR	.449	-.139	.339	.646
HipFLX	-.351	.067	-.740	-.140
HipABD	-.059	-.784	.142	.201
HipER	.095	-.191	.068	.858
FootAng	-.008	.223	.032	-.784
MKneFLX	.192	-.475	-.664	.256
MKneEXT	-.757	.190	.089	-.219
MKneABD	.900	.315	.084	-.038
MKneADD	.064	.951	.183	-.392
MKneIR	.075	-.381	-.084	.671
MKneER	-.653	.286	-.237	.199
MHipEXT	-.404	.124	.797	.068
MHipABD	.917	.112	.040	.143
MHipADD	.028	.922	.201	-.390
MHipIR	.807	.315	-.026	-.104
MHipER	.106	.858	.405	-.260

Extraction Method: Principal Component Analysis. , Rotation Method: Oblimin with Kaiser Normalization.

Component Correlation Matrix

Component	1	2	3	4
1	1.000	.020	.035	.108
2	.020	1.000	.102	-.232
3	.035	.102	1.000	-.006
4	.108	-.232	-.006	1.000

Extraction Method: Principal Component Analysis.
Rotation Method: Oblimin with Kaiser Normalization.

General Linear Model: FACTOR 1

Within-Subjects Factors

Measure: MEASURE_1

factor	Dependent Variable
1	FAC1RUN
2	FAC1LFS
3	FAC1SSC
4	FAC1LS

Between-Subjects Factors

		N
Sex	F	11
	M	10

Descriptive Statistics

	Sex	Mean	Std. Deviation	N
FAC1RUN	F	.1766	.67292	11
	M	.4348	.82682	10
	Total	.2996	.74264	21
FAC1LFS	F	.1709	.92448	11
	M	.3232	1.09406	10
	Total	.2434	.98593	21
FAC1SSC	F	-.1184	1.03296	11
	M	-.1407	1.18009	10
	Total	-.1291	1.07718	21
FAC1LS	F	-.2785	1.01611	11
	M	-.5629	1.13021	10
	Total	-.4140	1.05463	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Pillai's Trace	.547	6.834(b)	3.000	17.000	.003	.547	20.503	.937
	Wilks' Lambda	.453	6.834(b)	3.000	17.000	.003	.547	20.503	.937
	Hotelling's Trace	1.206	6.834(b)	3.000	17.000	.003	.547	20.503	.937
	Roy's Largest Root	1.206	6.834(b)	3.000	17.000	.003	.547	20.503	.937
factor * Sex	Pillai's Trace	.122	.787(b)	3.000	17.000	.517	.122	2.362	.184
	Wilks' Lambda	.878	.787(b)	3.000	17.000	.517	.122	2.362	.184
	Hotelling's Trace	.139	.787(b)	3.000	17.000	.517	.122	2.362	.184
	Roy's Largest Root	.139	.787(b)	3.000	17.000	.517	.122	2.362	.184

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+Sex
Within Subjects Design: factor

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
factor	.776	4.498	5	.481	.866	1.000	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+Sex
Within Subjects Design: factor

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Sphericity Assumed	7.298	3	2.433	7.146	.000	.273	21.437	.975
	Greenhouse-Geisser	7.298	2.597	2.811	7.146	.001	.273	18.554	.958
	Huynh-Feldt	7.298	3.000	2.433	7.146	.000	.273	21.437	.975
	Lower-bound	7.298	1.000	7.298	7.146	.015	.273	7.146	.718
factor * Sex	Sphericity Assumed	.883	3	.294	.865	.465	.044	2.594	.227
	Greenhouse-Geisser	.883	2.597	.340	.865	.452	.044	2.245	.211
	Huynh-Feldt	.883	3.000	.294	.865	.465	.044	2.594	.227
	Lower-bound	.883	1.000	.883	.865	.364	.044	.865	.143
Error(factor)	Sphericity Assumed	19.406	57	.340					
	Greenhouse-Geisser	19.406	49.335	.393					
	Huynh-Feldt	19.406	57.000	.340					
	Lower-bound	19.406	19.000	1.021					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	factor	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Linear	6.845	1	6.845	17.894	.000	.485	17.894	.980
	Quadratic	.283	1	.283	.724	.405	.037	.724	.128
	Cubic	.171	1	.171	.688	.417	.035	.688	.124
factor * Sex	Linear	.851	1	.851	2.225	.152	.105	2.225	.294
	Quadratic	.032	1	.032	.082	.778	.004	.082	.059
	Cubic	9.02E-005	1	9.02E-005	.000	.985	.000	.000	.050
Error(factor)	Linear	7.267	19	.382					
	Quadratic	7.427	19	.391					
	Cubic	4.711	19	.248					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	3.20E-005	1	3.20E-005	.000	.997	.000	.000	.050
Sex	.014	1	.014	.005	.945	.000	.005	.050
Error	55.619	19	2.927					

a Computed using alpha = .05

Estimated Marginal Means

1. Grand Mean

Measure: MEASURE_1

Mean	Std. Error	95% Confidence Interval	
		Lower Bound	Upper Bound
.001	.187	-.391	.392

2. Sex

Measure: MEASURE_1

Sex	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
F	-.012	.258	-.552	.527
M	.014	.271	-.553	.580

3. factor

Measure: MEASURE_1

factor	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	.306	.164	-.037	.649
2	.247	.220	-.214	.708
3	-.130	.241	-.635	.376
4	-.421	.234	-.911	.069

4. Sex * factor

Measure: MEASURE_1

Sex	factor	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
F	1	.177	.226	-.297	.650
	2	.171	.304	-.465	.807
	3	-.118	.333	-.816	.579
	4	-.279	.323	-.955	.398
M	1	.435	.237	-.061	.931
	2	.323	.319	-.344	.991
	3	-.141	.349	-.872	.591
	4	-.563	.339	-1.272	.146

General Linear Model: FACTOR 2

Within-Subjects Factors

Measure: MEASURE_1

factor	Dependent Variable
1	FAC2RUN
2	FAC2LFS
3	FAC2SSC
4	FAC2LS

Between-Subjects Factors

		N
Sex	F	11
	M	10

Descriptive Statistics

	Sex	Mean	Std. Deviation	N
FAC2RUN	F	1.2603	.24247	11
	M	.9047	.23327	10
	Total	1.0910	.29495	21
FAC2LFS	F	.2631	.45405	11
	M	.1636	.27111	10
	Total	.2157	.37249	21
FAC2SSC	F	.1011	.65218	11
	M	-.1501	.34834	10
	Total	-.0185	.53272	21
FAC2LS	F	-1.1037	.90502	11
	M	-1.4912	.57747	10
	Total	-1.2882	.77391	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Pillai's Trace	.929	74.016(b)	3.000	17.000	.000	.929	222.047	1.000
	Wilks' Lambda	.071	74.016(b)	3.000	17.000	.000	.929	222.047	1.000
	Hotelling's Trace	13.062	74.016(b)	3.000	17.000	.000	.929	222.047	1.000
	Roy's Largest Root	13.062	74.016(b)	3.000	17.000	.000	.929	222.047	1.000
factor * Sex	Pillai's Trace	.123	.792(b)	3.000	17.000	.515	.123	2.375	.185
	Wilks' Lambda	.877	.792(b)	3.000	17.000	.515	.123	2.375	.185
	Hotelling's Trace	.140	.792(b)	3.000	17.000	.515	.123	2.375	.185
	Roy's Largest Root	.140	.792(b)	3.000	17.000	.515	.123	2.375	.185

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+Sex
Within Subjects Design: factor

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
factor	.483	12.915	5	.024	.689	.815	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+Sex
Within Subjects Design: factor

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Sphericity Assumed	60.788	3	20.263	108.471	.000	.851	325.414	1.000
	Greenhouse-Geisser	60.788	2.067	29.404	108.471	.000	.851	224.248	1.000
	Huynh-Feldt	60.788	2.446	24.854	108.471	.000	.851	265.302	1.000
	Lower-bound	60.788	1.000	60.788	108.471	.000	.851	108.471	1.000
factor * Sex	Sphericity Assumed	.265	3	.088	.472	.703	.024	1.417	.140
	Greenhouse-Geisser	.265	2.067	.128	.472	.633	.024	.976	.123
	Huynh-Feldt	.265	2.446	.108	.472	.665	.024	1.155	.130
	Lower-bound	.265	1.000	.265	.472	.500	.024	.472	.100
Error(factor)	Sphericity Assumed	10.648	57	.187					
	Greenhouse-Geisser	10.648	39.280	.271					
	Huynh-Feldt	10.648	46.471	.229					
	Lower-bound	10.648	19.000	.560					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	factor	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Linear	57.024	1	57.024	166.996	.000	.898	166.996	1.000
	Quadratic	.854	1	.854	6.041	.024	.241	6.041	.645
	Cubic	2.909	1	2.909	37.516	.000	.664	37.516	1.000
factor * Sex	Linear	.016	1	.016	.047	.831	.002	.047	.055
	Quadratic	.202	1	.202	1.427	.247	.070	1.427	.206
	Cubic	.047	1	.047	.604	.447	.031	.604	.114
Error(factor)	Linear	6.488	19	.341					
	Quadratic	2.686	19	.141					
	Cubic	1.473	19	.078					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	.004	1	.004	.007	.934	.000	.007	.051
Sex	1.567	1	1.567	3.072	.096	.139	3.072	.384
Error	9.690	19	.510					

a Computed using alpha = .05

Estimated Marginal Means

1. Grand Mean

Measure: MEASURE_1

Mean	Std. Error	95% Confidence Interval	
		Lower Bound	Upper Bound
-.007	.078	-.170	.157

2. Sex

Measure: MEASURE_1

Sex	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
F	.130	.108	-.095	.356
M	-.143	.113	-.380	.093

3. factor

Measure: MEASURE_1

factor	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	1.083	.052	.974	1.191
2	.213	.083	.040	.386
3	-.024	.116	-.267	.218
4	-1.297	.168	-1.648	-.947

4. Sex * factor

Measure: MEASURE_1

Sex	factor	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
F	1	1.260	.072	1.110	1.411
	2	.263	.114	.024	.502
	3	.101	.160	-.234	.436
	4	-1.104	.231	-1.588	-.619
M	1	.905	.075	.747	1.062
	2	.164	.120	-.087	.414
	3	-.150	.168	-.501	.201
	4	-1.491	.243	-1.999	-.983

General Linear Model: FACTOR 3

Within-Subjects Factors

Measure: MEASURE_1

factor	Dependent Variable
1	FAC3RUN
2	FAC3LFS
3	FAC3SSC
4	FAC3LS

Between-Subjects Factors

		N
Sex	F	11
	M	10

Descriptive Statistics

	Sex	Mean	Std. Deviation	N
FAC3RUN	F	-.0205	.38569	11
	M	-.2455	.45425	10
	Total	-.1276	.42485	21
FAC3LFS	F	.6405	.73255	11
	M	.2097	1.02727	10
	Total	.4354	.88983	21
FAC3SSC	F	.0704	.57998	11
	M	-.7982	1.04631	10
	Total	-.3432	.92652	21
FAC3LS	F	.4801	.92150	11
	M	-.4537	1.71603	10
	Total	.0355	1.40645	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Pillai's Trace	.633	9.778(b)	3.000	17.000	.001	.633	29.335	.989
	Wilks' Lambda	.367	9.778(b)	3.000	17.000	.001	.633	29.335	.989
	Hotelling's Trace	1.726	9.778(b)	3.000	17.000	.001	.633	29.335	.989
	Roy's Largest Root	1.726	9.778(b)	3.000	17.000	.001	.633	29.335	.989
factor * Sex	Pillai's Trace	.240	1.787(b)	3.000	17.000	.188	.240	5.361	.382
	Wilks' Lambda	.760	1.787(b)	3.000	17.000	.188	.240	5.361	.382
	Hotelling's Trace	.315	1.787(b)	3.000	17.000	.188	.240	5.361	.382
	Roy's Largest Root	.315	1.787(b)	3.000	17.000	.188	.240	5.361	.382

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+Sex
Within Subjects Design: factor

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
factor	.376	17.314	5	.004	.619	.720	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+Sex
Within Subjects Design: factor

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Sphericity Assumed	6.918	3	2.306	7.032	.000	.270	21.097	.973
	Greenhouse-Geisser	6.918	1.858	3.724	7.032	.003	.270	13.063	.891
	Huynh-Feldt	6.918	2.159	3.204	7.032	.002	.270	15.183	.924
	Lower-bound	6.918	1.000	6.918	7.032	.016	.270	7.032	.711
factor * Sex	Sphericity Assumed	1.844	3	.615	1.874	.144	.090	5.622	.461
	Greenhouse-Geisser	1.844	1.858	.992	1.874	.171	.090	3.481	.351
	Huynh-Feldt	1.844	2.159	.854	1.874	.164	.090	4.046	.382
	Lower-bound	1.844	1.000	1.844	1.874	.187	.090	1.874	.255
Error(factor)	Sphericity Assumed	18.691	57	.328					
	Greenhouse-Geisser	18.691	35.295	.530					
	Huynh-Feldt	18.691	41.021	.456					
	Lower-bound	18.691	19.000	.984					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	factor	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Linear	.129	1	.129	.221	.644	.011	.221	.073
	Quadratic	.172	1	.172	1.719	.205	.083	1.719	.238
	Cubic	6.618	1	6.618	21.908	.000	.536	21.908	.993
factor * Sex	Linear	1.722	1	1.722	2.959	.102	.135	2.959	.372
	Quadratic	.026	1	.026	.260	.616	.013	.260	.077
	Cubic	.096	1	.096	.317	.580	.016	.317	.083
Error(factor)	Linear	11.056	19	.582					
	Quadratic	1.896	19	.100					
	Cubic	5.739	19	.302					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	.018	1	.018	.007	.934	.000	.007	.051
Sex	7.913	1	7.913	3.150	.092	.142	3.150	.392
Error	47.728	19	2.512					

a Computed using alpha = .05

Estimated Marginal Means

1. Grand Mean

Measure: MEASURE_1

Mean	Std. Error	95% Confidence Interval	
		Lower Bound	Upper Bound
-.015	.173	-.377	.348

2. Sex

Measure: MEASURE_1

Sex	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
F	.293	.239	-.207	.793
M	-.322	.251	-.846	.203

3. factor

Measure: MEASURE_1

factor	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-.133	.092	-.325	.059
2	.425	.193	.021	.830
3	-.364	.182	-.745	.017
4	.013	.296	-.607	.634

4. Sex * factor

Measure: MEASURE_1

Sex	factor	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
F	1	-.020	.127	-.285	.244
	2	.641	.267	.082	1.199
	3	.070	.251	-.456	.597
	4	.480	.409	-.376	1.337
M	1	-.245	.133	-.523	.032
	2	.210	.280	-.376	.795
	3	-.798	.264	-1.350	-.246
	4	-.454	.429	-1.352	.445

General Linear Model: FACTOR 4

Within-Subjects Factors

Measure: MEASURE_1

factor	Dependent Variable
1	FAC4RUN
2	FAC4LFS
3	FAC4SSC
4	FAC4LS

Between-Subjects Factors

		N
Sex	F	11
	M	10

Descriptive Statistics

	Sex	Mean	Std. Deviation	N
FAC4RUN	F	-.2677	.81951	11
	M	-.3342	.59679	10
	Total	-.2993	.70515	21
FAC4LFS	F	-.4830	.99889	11
	M	-.5851	.60846	10
	Total	-.5316	.81745	21
FAC4SSC	F	.4236	1.03026	11
	M	.6856	1.02038	10
	Total	.5484	1.00858	21
FAC4LS	F	.1663	1.06980	11
	M	.4106	1.14409	10
	Total	.2826	1.08485	21

Multivariate Tests(c)

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Pillai's Trace	.641	10.118(b)	3.000	17.000	.000	.641	30.355	.991
	Wilks' Lambda	.359	10.118(b)	3.000	17.000	.000	.641	30.355	.991
	Hotelling's Trace	1.786	10.118(b)	3.000	17.000	.000	.641	30.355	.991
	Roy's Largest Root	1.786	10.118(b)	3.000	17.000	.000	.641	30.355	.991
factor * Sex	Pillai's Trace	.049	.291(b)	3.000	17.000	.832	.049	.872	.095
	Wilks' Lambda	.951	.291(b)	3.000	17.000	.832	.049	.872	.095
	Hotelling's Trace	.051	.291(b)	3.000	17.000	.832	.049	.872	.095
	Roy's Largest Root	.051	.291(b)	3.000	17.000	.832	.049	.872	.095

a Computed using alpha = .05
b Exact statistic
c Design: Intercept+Sex
Within Subjects Design: factor

Mauchly's Test of Sphericity(b)

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon(a)		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
factor	.382	17.075	5	.004	.666	.783	.333

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.
a May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.
b Design: Intercept+Sex
Within Subjects Design: factor

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Sphericity Assumed	16.060	3	5.353	11.894	.000	.385	35.682	.999
	Greenhouse-Geisser	16.060	1.998	8.039	11.894	.000	.385	23.762	.991
	Huynh-Feldt	16.060	2.350	6.834	11.894	.000	.385	27.950	.996
	Lower-bound	16.060	1.000	16.060	11.894	.003	.385	11.894	.905
factor * Sex	Sphericity Assumed	.600	3	.200	.445	.722	.023	1.334	.134
	Greenhouse-Geisser	.600	1.998	.301	.445	.644	.023	.888	.117
	Huynh-Feldt	.600	2.350	.256	.445	.675	.023	1.045	.123
	Lower-bound	.600	1.000	.600	.445	.513	.023	.445	.097
Error(factor)	Sphericity Assumed	25.655	57	.450					
	Greenhouse-Geisser	25.655	37.958	.676					
	Huynh-Feldt	25.655	44.649	.575					
	Lower-bound	25.655	19.000	1.350					

a Computed using alpha = .05

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	factor	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
factor	Linear	8.549	1	8.549	9.946	.005	.344	9.946	.849
	Quadratic	.006	1	.006	.030	.865	.002	.030	.053
	Cubic	7.505	1	7.505	25.061	.000	.569	25.061	.997
factor * Sex	Linear	.440	1	.440	.512	.483	.026	.512	.104
	Quadratic	.000	1	.000	.002	.963	.000	.002	.050
	Cubic	.160	1	.160	.534	.474	.027	.534	.107
Error(factor)	Linear	16.331	19	.860					
	Quadratic	3.634	19	.191					
	Cubic	5.690	19	.299					

a Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Intercept	.000	1	.000	.000	.990	.000	.000	.050
Sex	.149	1	.149	.070	.795	.004	.070	.057
Error	40.786	19	2.147					

a Computed using alpha = .05

Estimated Marginal Means

1. Grand Mean

Measure: MEASURE_1

Mean	Std. Error	95% Confidence Interval	
		Lower Bound	Upper Bound
.002	.160	-.333	.337

2. Sex

Measure: MEASURE_1

Sex	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
F	-.040	.221	-.503	.422
M	.044	.232	-.441	.529

3. factor

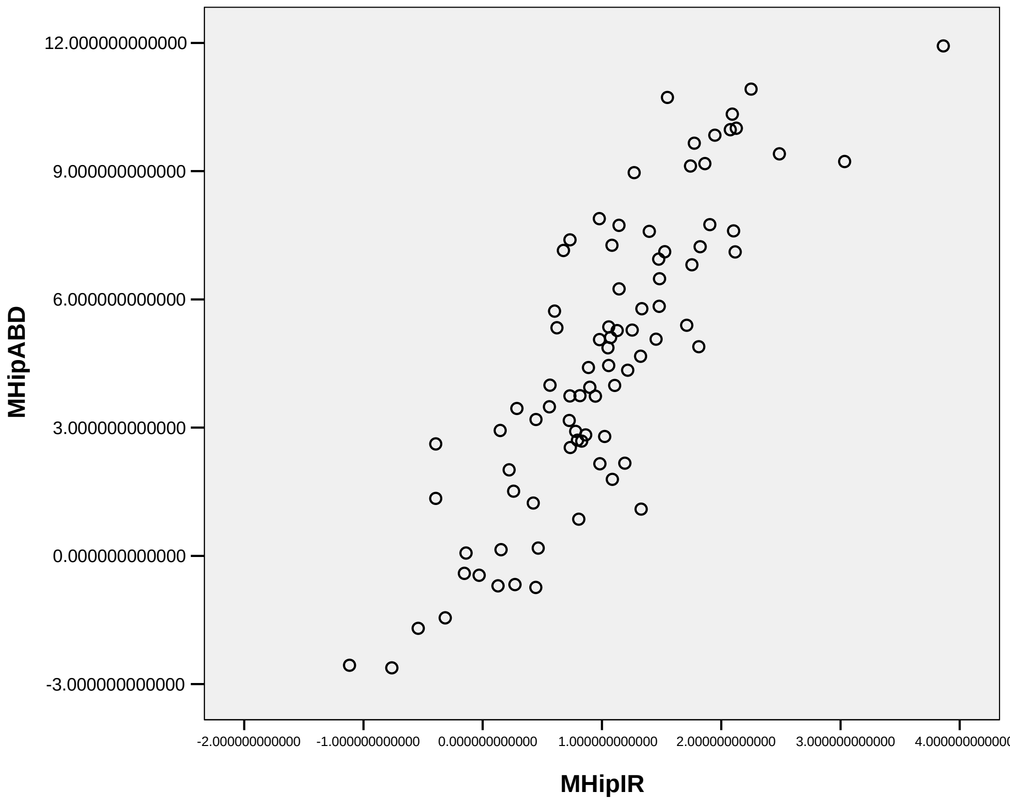
Measure: MEASURE_1

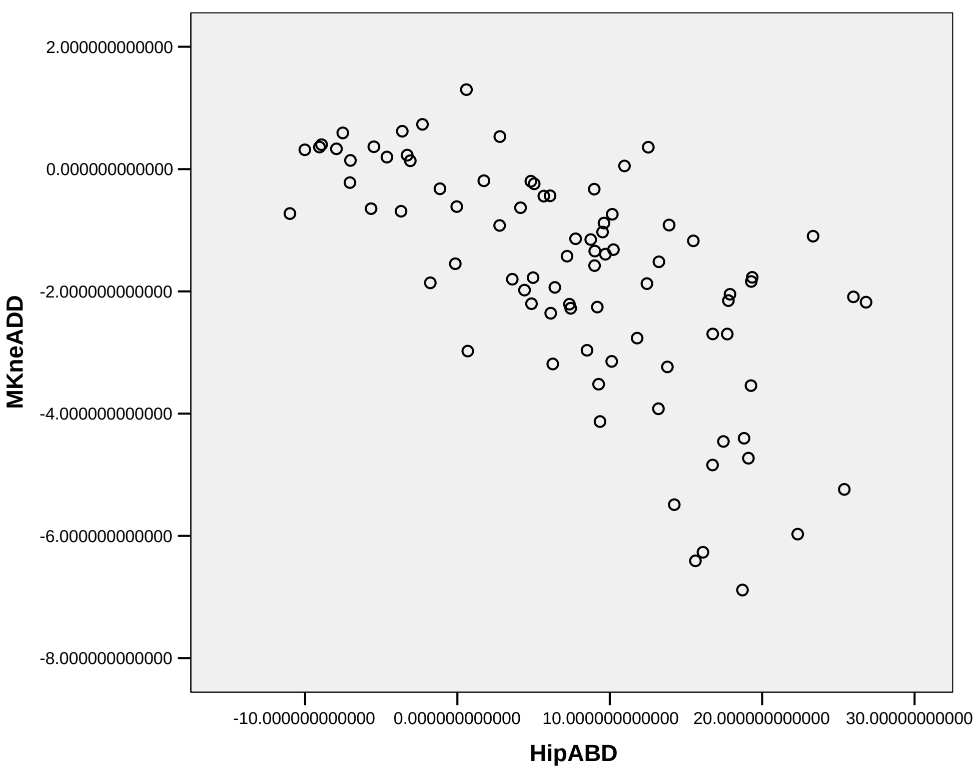
factor	Mean	Std. Error	95% Confidence Interval	
			Lower Bound	Upper Bound
1	-.301	.158	-.631	.030
2	-.534	.183	-.917	-.151
3	.555	.224	.086	1.024
4	.288	.242	-.217	.794

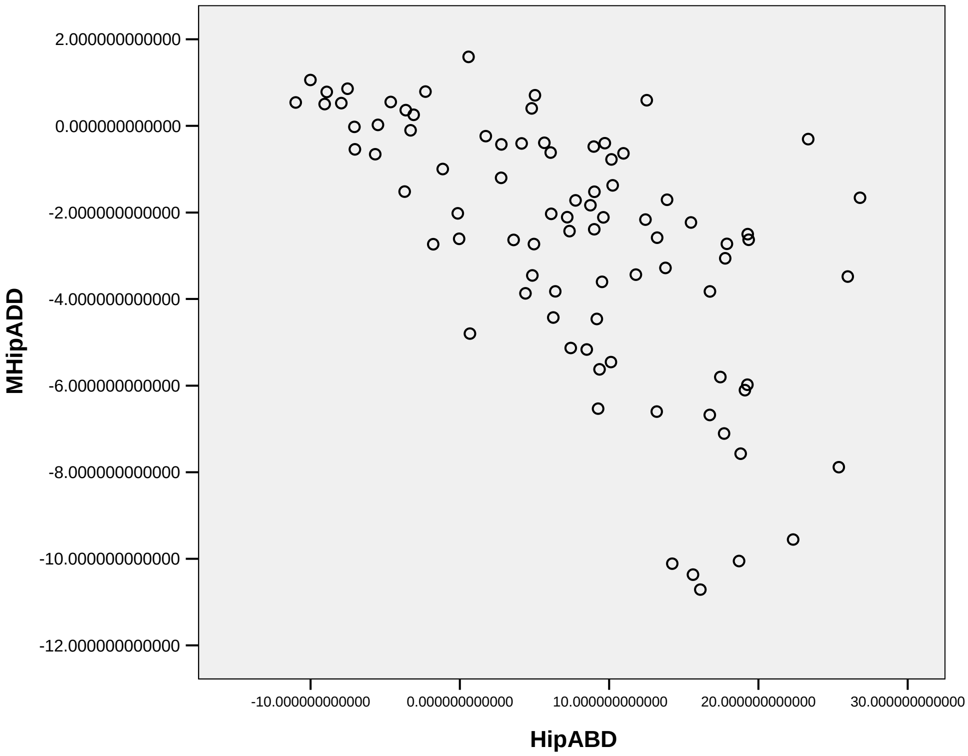
4. Sex * factor

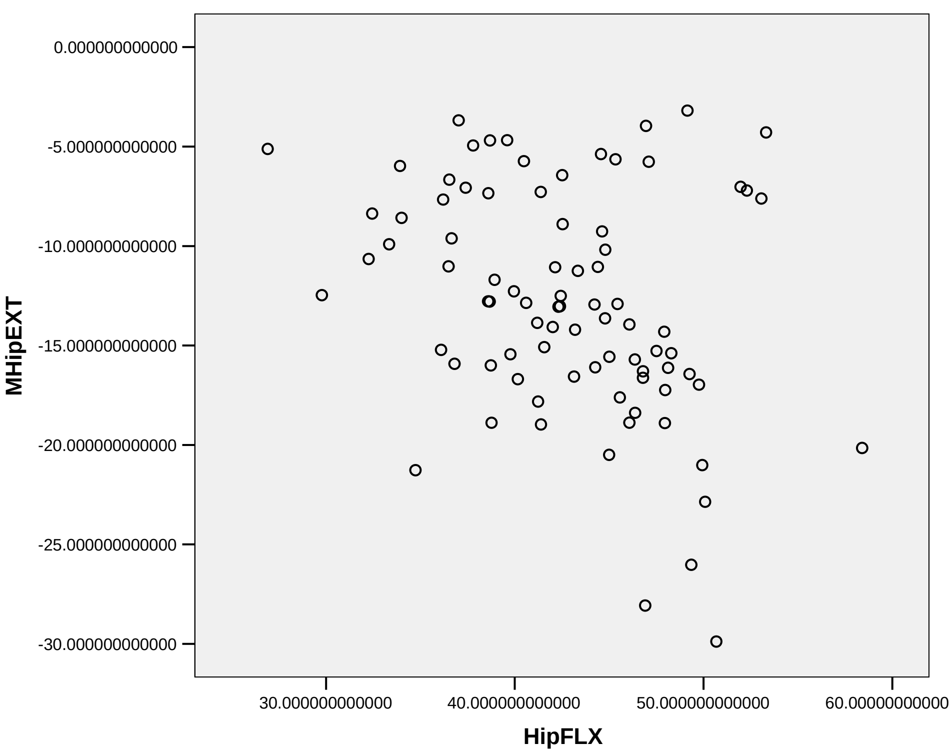
Measure: MEASURE_1

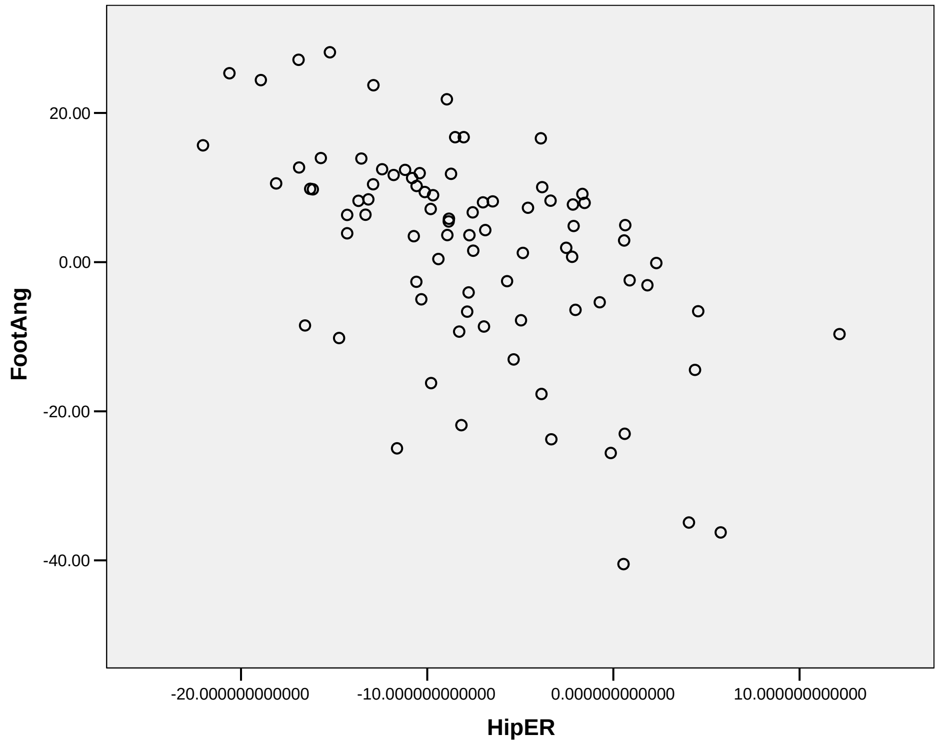
Sex	factor	Mean	Std. Error	95% Confidence Interval	
				Lower Bound	Upper Bound
F	1	-.268	.218	-.724	.188
	2	-.483	.252	-1.011	.045
	3	.424	.309	-.224	1.071
	4	.166	.333	-.531	.864
M	1	-.334	.229	-.812	.144
	2	-.585	.265	-1.139	-.031
	3	.686	.324	.007	1.364
	4	.411	.350	-.321	1.142

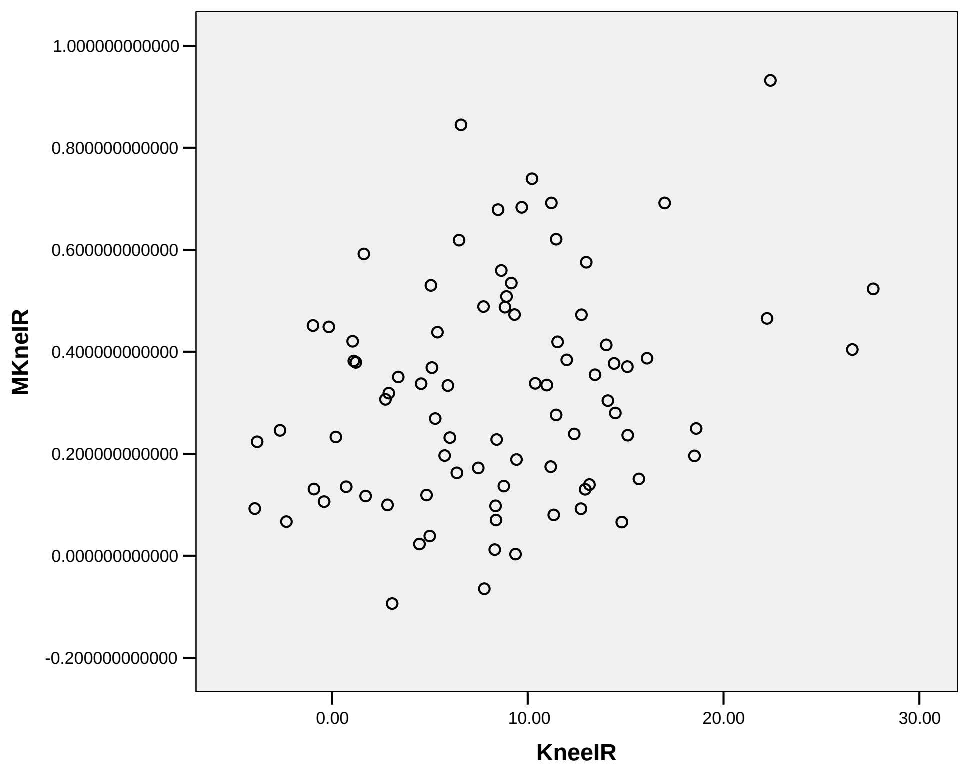

12.000000000000
9.000000000000
6.000000000000
3.000000000000
0.000000000000
-3.000000000000
MHipABD
-2.000000000000
-1.000000000000
0.000000000000
1.000000000000
2.000000000000
3.000000000000
4.000000000000
MHipIR


2.000000000000
0.000000000000
-2.000000000000
-4.000000000000
-6.000000000000
-8.000000000000
MKneADD
-10.000000000000
0.000000000000
10.000000000000
20.000000000000
30.000000000000
HipABD


2.000000000000
0.000000000000
-2.000000000000
-4.000000000000
-6.000000000000
-8.000000000000
-10.000000000000
-12.000000000000
MHipADD
-10.0000000000000
0.0000000000000
10.0000000000000
20.0000000000000
30.0000000000000
HipABD


0.000000000000
-5.000000000000
-10.000000000000
-15.000000000000
-20.000000000000
-25.000000000000
-30.000000000000
MHipEXT
30.0000000000000
40.0000000000000
50.0000000000000
60.00000000000
HipFLX


20.00
0.00
-20.00
-40.00
FootAng
-20.000000000000
-10.000000000000
0.000000000000
10.000000000000
HipER


1.000000000000
0.800000000000
0.600000000000
0.400000000000
0.200000000000
0.000000000000
-0.200000000000
MKneIR
0.00
10.00
20.00
30.00
KneeIR

Tukey's Honest Significant Difference

Modified Formula: accounting for unequal n

HSD = $q_T * \text{sqrt}((MS_{S/A}/2) / 1/n_1 + 1/n_2))$
$q_T = q_{0.05, 4, 21} = 3.96$
$MS_{S/A}$ = means square error from interaction for within subject effects
$n_1 = 11, n_2 = 10$

Variable	α (Condition)	$q_{0.05, 4, 21}$	$MS_{S/A}$	$MS_{S/A}$ / 2	$1/n_1 + 1/n_2$	HSD
Kinematics						
KA_ABD	0.000	3.96	0.481	0.2405	0.191	0.848
KA_IR	0.165	3.96	4.706	2.353	0.191	DNC
HA_ABD	0.000	3.96	1.274	0.637	0.191	1.381
HA_IR	0.000	3.96	9.183	4.5915	0.191	3.708
Kinetics						
KM_EXT	0.475	3.96	0.168	0.084	0.191	DNC
KM_ADD	0.000	3.96	0.532	0.266	0.191	0.892
KM_ER	0.251	3.96	0.043	0.0215	0.191	DNC
HM_ADD	0.000	3.96	1.547	0.7735	0.191	1.522
HM_ER	0.000	3.96	0.388	0.194	0.191	0.762
Factor Scores						
FAC1	0.000	3.96	0.294	0.147	0.191	0.663
FAC2	0.000	3.96	0.128	0.064	0.191	0.437
FAC3	0.003	3.96	0.992	0.496	0.191	1.218
FAC4	0.000	3.96	0.301	0.1505	0.191	0.671

Note: shaded areas indicate significance ($p < 0.05$)

Comparisons between group means

Variable	Condition	Mean	Comparison	ABS difference
KA_ABD	1	3.621	1v2	1.951
(0.848)	2	5.572	1v3	2.589
	3	6.21	1v4	3.444
	4	7.065	2v3	0.638
			2v4	1.493
			3v4	0.855
HA_ADD	1	-7.255	1v2	15.113
(1.381)	2	7.858	1v3	12.919
	3	5.664	1v4	23.879
	4	16.624	2v3	2.194
			2v4	8.766
			3v4	10.96
HA_IR	1	-4.15	1v2	1.03
(3.708)	2	-3.12	1v3	4.56
	3	0.41	1v4	5.18
	4	1.03	2v3	3.53
			2v4	4.15
			3v4	0.62

Comparison of group means (continued)

KM_ADD	1	0.13	1v2	1.29
(0.892)	2	-1.16	1v3	2.17
	3	-2.04	1v4	4.06
	4	-3.93	2v3	0.88
			2v4	2.77
			3v4	1.89
HM_ADD	1	0.014	1v2	1.792
(1.522)	2	-1.778	1v3	3.034
	3	-3.02	1v4	6.093
	4	-6.079	2v3	1.242
			2v4	4.301
			3v4	3.059
HM_ER	1	0.16	1v2	0.21
(0.762)	2	-0.05	1v3	0.72
	3	-0.56	1v4	1.47
	4	-1.31	2v3	0.51
			2v4	1.26
			3v4	0.75
FAC1	1	0.306	1v2	0.059
(0.663)	2	0.247	1v3	0.436
	3	-0.13	1v4	0.727
	4	-0.421	2v3	0.377
			2v4	0.668
			3v4	0.291
FAC2	1	-1.083	1v2	0.87
(0.437)	2	-0.213	1v3	1.107
	3	0.024	1v4	2.38
	4	1.297	2v3	0.237
			2v4	1.51
			3v4	1.273
FAC3	1	0.133	1v2	0.558
(1.218)	2	-0.425	1v3	0.231
	3	0.364	1v4	0.146
	4	-0.013	2v3	0.789
			2v4	0.412
			3v4	0.377
FAC4	1	0.301	1v2	0.233
(0.671)	2	0.534	1v3	0.856
	3	-0.555	1v4	0.589
	4	-0.288	2v3	1.089
			2v4	0.822
			3v4	0.267

Note: shaded areas indicate significance ($p < 0.05$)