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# Effect of Frontal Plane Foot Position on Lower Extremity Muscle Activation and Limb Positioning in a Single Leg Squat

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# EFFECT OF FRONTAL PLANE FOOT POSITION ON LOWER EXTREMITY MUSCLE ACTIVATION AND LIMB POSITIONING IN A SINGLE LEG SQUAT

by

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This Scholarly Project, submitted by Marissa N. Laddusaw in partial fulfillment of the Requirements for the Degree of Doctor of Physical Therapy from the University of North Dakota, has been read by the Advisor and Chairperson of Physical Therapy under whom the work has been done and is hereby approved.

____________________________________
(Graduate School Advisor)

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ABSTRACT

*Purpose/Background:* This goal of this study was to reach a bottom up understanding of ACL injury occurance. Subtalar position during single leg stance may affect the muscles associated with the knee joint and promote ACL injury. The muscle activity of six muscles of the leg, thigh, and hip were analyzed during a single leg squat with the foot on surfaces of 10 degrees decline, five degrees decline, neutral, and 5 degrees incline, and 10 degrees incline.

*Methods:* Seventeen healthy participants between the ages of 18 and 30 performed five rhythmically timed repetitions of single leg squats on surfaces of 10 degrees decline, five degrees decline, neutral, and 5 degrees incline, and 10 degrees incline to simulate pronation with declined surfaces and supination with inclined surfaces. Electrodes were placed over selected muscles and data on their activity was collected and evaluated.

*Results:* Muscle activity increased with increasing inclination and declination. Percent MVC increased the most in ten degrees of subtalar pronation in four of six tested muscles: anterior tibialis, lateral gastrocnemius, gluteus maximus and gluteus medius. However, due to study limitations, high variability, and large standard deviations we are unable to make a confident conclusion. Lateral gastrocnemius, biceps femoris and gluteus maximus were found to be statistically significance using Friedman's.

*Conclusions:* A pronated subtalar position results in the highest %MVC in the majority of muscles of the lower extremity proximal to the ankle joint. This suggests that subtalar position affects muscles and joints more proximal which may result in an increased risk of ACL injury in this position. More research is needed to determine the effect subtalar position on susceptibility to ACL injury as our study has many limitations.

# CHAPTER I

# INTRODUCTION

The anterior cruciate ligament (ACL) has been a highly researched topic due to the frequent occurrence of ACL injuries. Approximately 200,000 ACL injuries occur in the United States annually.[1] Costs of surgery and therapy following an ACL injury can range from $17,000 to $25,000.[2,3] Injury may result in loss of participation of an entire sports season, loss of scholarship awards, and decreased scholastic performance.[4] Problems continue to arise after an ACL tear. There is an increased risk of reinjury to the affected knee and injury to the unaffected knee. Osteoarthritic changes are often seen after an ACL tear (T.E. Hewett, T. Malone personal communication, June 18, 2010).

Cause of ACL injuries and the difference in occurrence between genders is likely caused by many factors acting simultaneously. These same factors are thought to influence other pathologies of the knee.[5] Females have been found to be between 4 and 8 times more likely to sustain an ACL injury compared to males.[6-11] There have been many factors suggested to contribute to noncontact ACL tears. About 70% of ACL tears are due to noncontact injuries.[12] ACL injuries occur in females most frequently during noncontact movements such as cutting, landing, and decelerating.[13]

The ACL attaches from the posteromedial boarder of the lateral femoral condyle to the anterior tibial spine.[14] The ACL is made up of three bundles; anteromedial bundle, posterolateral bundle, and intermediate bundle.[15] The anteromedial bundle provides anterolateral stability as tension increases with knee flexion. The posterolateral bundle contributes posterolateral stability as tension increases with knee extension. Anterior and anteromedial stability are contributed by the intermediate bundle.[14] There are six degrees of freedom described as motions in relation to the tibia on the femur: flexion and extension, adduction and abduction, internal and external rotation, anterior and posterior translation, medial and lateral translation, and compression and distraction. The ACL resists anterior translation, varus, valgus, knee hyperextension, and internal rotation (T.E. Hewett, T. Malone personal communication, June 18, 2010).[15] It is approximately the size of the 5th finger and is 5% nervous tissue.

# CHAPTER II

# LITERATURE REVIEW

While it has been highly recognized that there are gender differences in the occurrence of ACL injuries, it is unknown exactly why females are at greater risk for sustaining an ACL injury. The most commonly identified factors include trunk dominance, quadriceps dominance, ligament dominance, and leg dominance.[16] Neuromuscular imbalance, foot structure, knee joint structure, and hormones are included in the growing list of other factors believed to influence ACL injury.[5] Injury to the ACL occurs with typical posturing of the knee and trunk during deceleration, pivoting, or landing under high loads (T.E. Hewett, T. Malone personal communication, June 18, 2010).[16] Knee alignment during landing is a result of the contributing factors listed. Females land with greater knee abduction angles than men from various heights and landing techniques which may increase stress on the ACL.[17]

## Neuromuscular Control

Neuromuscular control can be defined as having adequate muscle strength, power, and activation pattern.[18] Due to the typical demonstration of decreased neuromuscular control in females, greater loads are placed on the lower extremities during dynamic activities, resulting in an increased risk of ACL injury. Inadequate neuromuscular control may negatively affect the position of the knee joint and place the ACL at an increased risk of injury. [19]

## Trunk Dominance

Trunk dominance is defined as the inability of core muscles associated with the trunk and hips to match the inertia of the trunk during dynamic movements.[16] The core muscles have been found to play a large part in maintaining correct lower extremity (LE) alignment during dynamic movements.[20] When the core muscles fire prior to distal muscles of the extremities, proper alignment is better maintained.[21] When core muscles do not fire prior to distal muscles, the trunk is allowed to move more laterally which increases forces into knee abduction.[22] Adequate control and coordination of the trunk muscles are necessary to control moments of inertia. Movements of the female athlete are determined more by the trunk's inertia than by adequate functioning of the core muscles.[16] This results in excessive movement of the trunk causing increased ground reaction forces (GRF) and valgus forces into the LEs.

Inability to maintain the center of mass (COM) over the base of support (BOS) is highly correlated with poor core stability. The GRF is directed up to the center of mass (COM) located at approximately S2. Because the location of COM is located in the spine, trunk control is important to maintain the torso over the BOS. With poor trunk control, the COM moves outside of the BOS causing irregular forces through the body. Lateral movement of the COM outside of the BOS causes the knee to move into a valgus position, a component of common body positioning during ACL tears. Inability to control the trunk three dimensionally has not been observed in men at the time of ACL injury.[5] Zazulak et al[23] found that core stability was a factor in predicting ACL injury risk in females, but did not find this to be true for males. Neuromuscular control at the hip has been show to affect the amount of knee valgus that occurs with dynamic movements.

## Hip Neuromuscular Control

Zazulak et al demonstrated electromyographic differences between males and females in firing patterns of muscles associated with the hip joint. During single-leg landing, females demonstrated an increased firing of the quadriceps, a decreased firing of the gluteal muscles compared to men, and an increase in hip adduction angles and torques.[24] It is likely that an increase in the adduction angle contributes to an increase in knee valgus during dynamic movements, placing an individual at increased risk for ACL injury.[25]

If hip extensors and abductors are unable to react to ground reaction forces then other joints more distal in the lower extremities must compensate for this loss. Weak abductors result in an unbalanced pelvis in the frontal plane and may cause the femur to internally rotate and adduct.[26] This internally rotated and adducted position of the femur can cause changes in more distal joints. Individuals with pathologies at the knee have been shown to have poor neuromuscular control at the hip joint.[27] If alterations in hip neuromuscular control continue, alternative undesirable compensatory patterns may form and one or more joints may be subjected to mechanical overload.[28]

## Quadriceps Dominance

Coactivation of the knee flexors and extensors has been deemed important for knee stabilization.[29] Quadriceps dominance refers to the imbalance between the quadriceps muscle group and the hamstring muscle group in terms of neuromuscular control, strength, and coordination.[30] It has been found that men “turn on” their hamstrings first while the quadriceps engage first in females (T.E. Hewett, T. Malone personal communication, June 18, 2010). This causes females to rely on the quadriceps muscle group for stabilization of the knee joint particularly during dynamic activities.[5] Hamstring weakness in relation to quadricep strength is believed to increase the risk of injuries in the LEs[31] as well as injuries to the ACL.[30] Hewett et al[32] found that females land with a decreased knee flexor torque than men. Females in this study also demonstrated a greater difference from right to left hamstring peak torque compared to male participants. Ford et al[33] found that compared to males, females land with a larger maximum valgus knee angle and a larger total valgus knee motion. Females in this study were also found to demonstrate differences between their dominant and nondominant legs in terms of maximum knee valgus angle.

ACL injuries commonly occur with the knee near full extension, due in part to the action of the quadriceps muscle group.[34] The pull of the quadriceps on the tibia via the patellar tendon introduces a forward translation of the tibia and an anterior shear force on the knee joint. Anterior shear forces in the knee are resisted by the ACL. Therefore, unbalanced engagement of the quadriceps compared to the hamstrings induces a tensile force on the ACL and places the ACL in a position of increased risk of injury.[5] This unequal activation of the hamstrings and quadriceps may negatively affect females' abilities to stabilize the knee joint during dynamic activities.[16]

Cocontraction of the hamstrings with the quadriceps as opposed to quadriceps dominance stabilizes the knee joint and may decrease the stress placed on passive structures,[35] increases knee joint compression force, and stabilizes the knee in neutral varus/valgus.[36] ACL strain was found by Withrow et al to be decreased with an increase in hamstring force during the flexion phase of a jump landing.[37] Hewett et al. found that compared to men, women demonstrated a flexion moment that was three times less than that of men during a landing, a decreased isokinetic hamstring to quadriceps ratio, and an increase in knee valgus moments.[32]

The hamstrings play an especially vital role in stabilizing the knee joint due the muscles' attachment sites. The semitendinosus and semimembranosus attach on the medial side of the tibia, and the biceps femoris long head and short head attach to the lateral side of the fibula. These muscles provide stability and support on either side of the knee joint. Their attachment sites also allow the muscles to provide an effective check on the action of the quadriceps muscle group. The hamstrings act to move the knee into flexion. With co-contraction of both muscle groups, the knee is allowed to assume a more flexed position during landing. In this position, the muscles are more efficient shock absorbers. Adequate hamstring engagement ensures that less stress is placed on the ACL as the hamstrings oppose the quadriceps and provide an opposing posterior shear force.[5]

In full extension, the horizontal force of the quadriceps on the tibia are greatest, resulting in a greater anterior shear force of the tibia on the femur.  As the knee moves into flexion, the patellar tendon comes to a less advantageous position to exert a horizontal force on the tibial tuberosity where it inserts.  Hence, this results in a decrease in the anterior shear force resulting from the quadriceps.[38]  Essentially, quadriceps dominance in women results in an inability to adequately co-contract the hamstrings to prevent the occurrence of potentially harmful anterior shear force in the knee.  Athletes who land with a low knee flexion angle and a pronated foot are at an increased risk of ACL injury and typically demonstrate quadriceps dominance.[16]

In the study conducted by Huston and Wojtys,[4] the tibia was pulled forward by a turnbuckle device.  They found that men engaged their hamstrings first whereas women engaged their quadriceps first.  Podraza and White[39] demonstrated that knee extensor torque increased and ground reaction forces decreased with more knee flexion, but while knee extensor torque decreased with knee extension, ground reaction forces were greatest in this position.  This study thus concluded that anterior shear force imparted on the tibia by the quadriceps may not be the main cause for ACL injuries.

## Dynamic ACL Antagonists of the Leg

Research addressing neuromuscular influence typically discusses the muscles associated with the thigh, leaving out the leg muscles.[39,40]  The gastrocnemius can exert a posterior shear force of the femur on the tibia resulting in an anterior translation of the tibia on the femur as with quadriceps engagement.  This is the action resisted by the ACL.  However, the soleus, is believed to impart a posterior pull of the tibia on the femur reducing the anterior shear force the ACL must work against; therefore, the soleus acts as an agonist to the ACL.[41]

## Ligament Dominance

Ligament dominance is the condition in which ligaments, bony structures, and articular cartilage are forced to absorb shock due to lack of muscle control.  In other words, stabilization of the knee joint during dynamic activities is accomplished through an imbalance between neuromuscular and ligamentous control.[30] This imbalance is demonstrated by the inability to control the knee in the frontal plane during such activities. Ligament dominance is related to quadriceps dominance due to the lack of posterior chain muscles' ability to engage adequately and provide dynamic stabilization. The muscles of the posterior leg and thigh are particularly important to prevent ligament dominance from occurring in the knee joint. These muscles include gluteus maximus and minimus, the hamstring muscle group, and soleus.[5]

The GRF is particularly harmful when absorbed through non-contractile tissues. For every action there is an equal and opposite reaction, according to Newton's third law of motion. However, the GRF is amplified through the body due to the velocity of moving body parts. In the case of ligament dominance, the result becomes a multiplied force absorbed by non-contractile tissues, such as the ACL. The ACL and other non-contractile tissues of the knee are placed at an increased risk of being injured, particularly if the force is strong enough to cause tissue failure.[5]

## Leg Dominance

Leg dominance refers to an imbalance between the right and left LEs in terms of strength, coordination, and neuromuscular control. Favoring one leg results in an increased frequency in weight bearing on the dominant leg, increasing its likelihood of sustaining an injury. At the moment of an ACL tear, a female is typically bearing all or the majority of her weight on one leg. While nearly all athletes have a preferential leg for planting, kicking, or jumping, there appears to be a greater difference from side to side in females than males. During a jump descent, the dominant leg typically hits first and moves into a greater amount of valgus compared to the nondominant leg (T.E. Hewett, T. Malone personal communication, June 18, 2010).

## Hormonal Influence

Contradictions exist in regards to the relation of ligament laxity and ligament injury. One source suggests ligament laxity in women may contribute to knee instability during the menstrual cycle (T.E. Hewett, T. Malone personal communication, June 18, 2010). Ligament laxity may be a predisposing factor for ligament avulsion,[42] and ACL injury was more common in those with ligament laxity in another study[44]; however, other studies found no correlation between ligament laxity and ligament injury.[44.45]

Estrogen is believed to have a direct effect on tendon compliance and neuromuscular control, particularly in females. There are estrogen receptors located on skeletal muscle and therefore may influence neuromuscular control at the tissue level.[46] Estrogen receptors are also found throughout the central nervous system (CNS) which may alter input to effector organs. The presence of estrogen causes a decreased production of collagen in tendons possibly by interacting with estrogen receptors found on the ACL. This is done by altering fibroblast activity. In other words, with increased estrogen blood levels there is also an increase in tendon compliance.[47]

## Landing Kinetics and Kinematics

Typical body position during noncontact ACL injuries has been noted to include knee abduction, single leg stance, knee extension, and with the COM away from the BOS (T.E. Hewett, T. Malone personal communication, June 18, 2010).[48] ACL injuries typically occur during dynamic movements when the dynamic stabilizers of the body are unable to control joint movements or dampen forces placed on joints.[49] Increased stress on ligaments occurs with decreased neuromuscular control of the surrounding muscles. If the passive stress on the ligaments exceeds their limit, the ligament may fail.[50]

Knee abduction results in the knee collapsing medially and places an increased tensile force on the ligaments of the knee. Females have been found to move into knee valgus while squatting in preparation to jump as well as during shock absorption after jumping. Compared to males, females land with less knee flexion and more knee valgus.[51,52] It is believed that a valgus knee position during active movements increases the stress placed on the ACL.[53] Knee valgus is referred to as a knock-kneed position.[16]

The knee is most often positioned in an extended position during the time of contact.[54] It has been hypothesized that this position, seen mainly in women, is due to quadriceps dominance, as differences in hamstring strength and neuromuscular control have been noted between males and females. Females land with more knee valgus from a variety of heights than do men. Stress on the ACL increases greatly when dynamic valgus moments are placed on the knee joint.[55]

## Developmental Differences

A difference is noted between mature men and women in the relationship between muscle strength and body size. During prepubescence, muscle strength compared to body weight is approximately the same ratio in males and females. However, after puberty males become disproportionately strong compared to their weight while females remain at the same prepubescent ratio. This has been demonstrated with vertical jumping, a demonstration of whole body power. In females, the vertical jump remains at a consistent ratio with body weight prior to and after puberty. In males, however, their vertical jump in ratio to body weight increases after puberty. Males experience increases in muscle power, strength, and coordination with maturation while these factors remain relatively constant in females.[56] Other changes noted in mature females include increased body fat, a change in the distribution of muscle mass, and a higher COM. In terms of core control, collectively this means men are better able to control their new larger body with their muscles of disproportionately greater strength. Women, on the other hand, are less able to control their larger body with muscles of unchanging, proportionate strength.[5] An increase in tibia length and mass that occurs with a growth spurt increases the amount of knee valgus during dynamic tasks and is associated with an increased risk of ACL injury.[16]

## Foot Structure

Foot structure has been explored as a factor of ACL tears. The amount of pronation may influence the susceptibility of sustaining an ACL injury, and differences between genders have been explored. A pronated foot in static standing has been shown to affect joints more proximal in the lower extremity and the pelvis.[57] In one study, females were found to have a greater navicular drop and larger Q-angles.[40] Another study compared navicular drop scores between men and women and injured and noninjured ACLs. The study found no significant difference between men and women. However, a significant difference was seen in navicular drop scores between individuals with injured ACLs compared to individuals with healthy ACLs.[58] An abnormal increase in navicular drop is believed to contribute to low back pain, hip pain, knee pain, and ankle pain.[59] Issues of biomechanics and alignment in the foot and ankle are known to influence the knee joint and propagate up the chain.[60] A study which included long distance runners found 77% of problems related to the knee were due to foot malalignment. Of those individuals, 43% demonstrated a pronated foot, and 34% were females with cavus alignment.[61]

Excessive pronation during the stance phase may alter the biomechanics of joints more proximal. Pronation and tibial internal rotation occur together.[62] During gait, as the knee flexes, the tibia internally rotates. Conversely, as the knee extends, the tibia externally rotates. Typically, tibia internal rotation and pronation occur simultaneously during the contact phase. However, with an excessively pronated foot, the tibia remains internally rotated longer than normal as the knee moves into flexion, hindering the tibia from moving into external rotation and the foot into supination as is normal in midstance.[58] During tibial internal rotation, tension on the ACL increases.[15] Tension decreases on the ACL during external rotation.

Femoral internal rotation, femoral adduction, tibial internal rotation, and pronation are all components of knee valgus angle.[63] The greater the amount of pronation, the greater the knee valgus angle and a greater anterior translation of the tibia on the femur. This results in an increased load on the ACL.[64] One study examined a group of female Division I athletes found that by limiting the amount of pronation in a single- leg drop jump, the knee valgus angle could be significantly reduced.[65] Another study compared navicular drop between individuals with ACL injury and individuals without ACL injury and found that navicular drop values were significantly higher in the ACL-deficient group.[66]

Individuals with a pronated foot that is acquired have been found to have decreased walking speed compared to normal.[67] It is possible that a flat foot may also cause movement dysfunction during other dynamic tasks. A pronated forefoot results in an everted hind foot,[27] decreased plantar flexion moment, and decreased push off force when walking.[68] A flat foot alters the ability of the calf muscles to produce force as is normal and manifests as relative weakness.[67] Malalignments at the foot may translate to alignment issues more proximal in the lower extremities and therefore affect the knee joint. Altered mechanical advantage of the posterior leg muscles might also translate to the muscles being disadvantaged in terms of stabilizing the knee joint. Kulig et al[67] found that compared to the control group, females with posterior tibialis tendon dysfunction demonstrated significant differences in plantar flexor strength and power, bilateral hip extensor strength and endurance, bilateral hip abductor strength, and distance walked during the six minute walk test. From these findings, it can be rationalized that changes in foot structure may coexist with decreased strength and endurance of muscles associated with the hip joint. Because muscles of the hip and ankle have direct and indirect effects on the knee joint, alterations in their functioning may influence susceptibility of ACL injury.

## Femoral and Tibial Alignment

Structure of the femur and tibia affect the alignment of the knee. A study compared measurements of the knee in ACL deficient individuals and unaffected individuals. The ACL-deficient group had significantly larger femoral plateau angles and tibial posterior slope angles.

The femoral plateau angle is the relation of the long axis of the femur to the superior plateau of the tibia. The tibial posterior slope 90 degrees minus a line made by the long axis of the tibia and the line made by the posterior plateau slope of the tibia. A decreased tibial slope results in hyperextension, and an increase in tibial slope results in a flexion deformity.[69] A large femoral plateau angle may be due to either an increased tibial plateau angle or knee hyperextension with a normal tibial plateau angle. In this study, female participants in the ACL deficient group showed an increased femoral plateau angle compared to female participants in the healthy ACL group. Because the anterior shear force of the quadriceps on the tibia is greatest at full extension, it can be implied that the anterior shear force on the tibia increases with a larger femoral plateau angle.[70]

## Identifying Individuals at Risk for ACL Injury

Activities associated with ACL injuries, for example landing, cutting, and decelerating, may be used to identify high-risk individuals with impaired neuromuscular dynamic control of the dynamic knee stabilizers.[16] It has been found that individuals at high risk for ACL injury will benefit more from neuromuscular training. The knee joints of these individuals are typically under higher loads during dynamic activities than their peers with adequate neuromuscular control.[71] Examining external knee abduction moment measurements at the time of landing may be helpful in identifying individuals at risk for ACL injury.[53] Factors useful in identifying individuals at risk for ACL injury include knee abduction moment, knee abduction angle, increased quadriceps recruitment compared to the hamstrings, decreased knee flexion range of motion, increased tibia length and mass.[53,54]

Myer et al[16] describe the tuck jump assessment as an effective tool for identifying individuals, specifically athletes, at a high risk of ACL injury. Ligament dominance is recognized with a landing in knee valgus and feet at a position narrower or wider than shoulder width. In other words, the athlete demonstrates an inability to control the LEs in the frontal plane. Individuals demonstrating quadriceps dominance land with flat feet and very little flexion of the knee and hip joints. This is also referred to as excessive landing contact noise. Leg dominance or residual injury deficits maybe be picked up by observing thighs at the height of the tuck jump, foot placement at landing, and foot contact with the floor timing. Increased risk of ACL injury occurs when the thighs are not level at the height of the tuck jump, when the feet land with one further ahead or behind the other, and when the feet do not contact the ground at the same time. Trunk dominance is identified when the thighs do not reach parallel by the peak of the jump, if the individual pauses between jumps, and/or if the individual is unable to land in the same spot. Tuck jumps are performed for 10 seconds, but the athlete is instructed to stop if technique declines significantly before the 10 seconds are up.

Subtalar position, namely hyperpronation, has been suggested to play a factor in ACL tears. The purpose of our study was to assess the relation between subtalar position during single leg squatting and electrical activity of select muscles of the leg, thigh, and hip by performing a single leg squat (SLS) on the dominant stance leg in a neutral foot position, 5° pronation, 10° pronation, 5° supination, and 10° supination. It is our hope to obtain a bottom-up perspective of the body in a single leg stance position of the dominant support leg. Following the acquisition of data, muscle activity will be evaluated in terms of motor control in each foot position.

# CHAPTER III

## MATERIALS AND METHODS

Prior to participation, test subjects were given written and verbal information on the purpose and methods of the study.  Consent to participation was obtained by signing a consent form.

A lower extremity history/measures profile form was filled out in order to take note of any surgical or injury history.  Gender was noted as was pregnancy status and the date on which the last menstrual cycle began in females.  Age in years, height in centimeters, and weight in kilograms were recorded (Table 1).  Males and females between the ages of 18 and 30 years were eligible for participation.  Participants were excluded if they had an acute lower extremity injury, were pregnant, less than 18 years of age, and older than thirty years of age.

| Table 1: Subject Data. | | | | |
|---|---|---|---|---|
| Subject | Age | Height (cm) | Weight (kg) | Dominant limb |
| 1 | 22 | 166 | 54.5 | L |
| 2 | 28 | 179 | 86.6 | L |
| 3 | 24 | 162 | 61.4 | L |
| 4 | 24 | 167.5 | 58.9 | L |
| 5 | 22 | 160 | 59.1 | L |
| 6 | 23 | 178 | 91 | L |
| 7 | 22 | 160.5 | 53 | L |
| 8 | 23 | 181 | 80.5 | L |
| 9 | 23 | 177 | 74 | L |
| 10 | 26 | 164.5 | 58 | R |
| Age, height, weight, and dominant leg of study participants. | | | | |

The dominant leg was determined by a ball kick. An examiner rolled a ball towards the subject and the subject spontaneously chose a stance leg and a kicking leg. The stance leg was deemed the dominant leg and was used throughout testing.[72] The length of the tested lower extremity was recorded in centimeters using the anterior superior iliac spine (ASIS) and the medial malleolus as the proximal and distal end points. EMG electrodes were then placed on the dominant, tested leg.

The following muscles were included in the EMG portion of our study: gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, gluteus medius, and gluteus maximus. Electrode sites were determined according to Criswell[73]. The gastrocnemius site was located one third of the distance from the fibular head to the calcaneus. The site for the tibialis anterior was placed one third of the distance from the fibular head to the calcaneus. The rectus femoris site was located halfway between the anterior superior iliac spine (ASIS) and the superior patella or a minimum of ten centimeters cephalad from the patella. Midway between the ischial tuberosity and the lateral femoral  condyle was the biceps femoris electrode site.  The gluteus medius site was found by determining half the distance between the most superior aspect of the iliac crest and the greater trochanter.  The gluteus maximus site was located half the distance between the inferior lateral angle of the sacrum and the greater trochanter.  Electrode sites were located above and below the markings just described with approximately 1 cm between them.  Impedance between electrodes was 50K or less.

Prior to electrode application, the electrode sites were prepared to allow optimal signal reading from the muscle to the computer.  Because hair can increase the impedance of the skin, an electric shaver was used to remove hair over the site if necessary.  Dead skin can also contribute to a weak signal from the muscle to the electrode.  To loosen dead tissues from the electrode site, the skin was lightly brushed with sandpaper, and lastly, the site was wiped with an alcohol soaked towel to remove all dead tissues.

Supination and pronation were simulated using inclined and declined surfaces respectively.  Single leg squats were performed barefoot on a level surface, on 5° of inclination, on 10° of inclination, on 5° of declination, and on 10° of declination. For each subject, a SLS on the level surface was performed first.  This reading was treated as the baseline muscle activity and was compared to the other degrees of supination and pronation.  Cards were drawn by the participant to randomly determine the order of SLS on the remaining surfaces.  This order was left untold to the participant until immediately prior to performing a SLS on each surface.  During each trial, the subjects lowered themselves by performing a SLS to approximately 55° knee flexion for 3 repetitions.  With each surface the subjects had one practice set with which to familiarize themselves.

Following completion of the SLS series, electrodes were removed.  Sites were inspected for adverse skin reactions and cleaned using a towel damp with rubbing alcohol.

## CHAPTER IV

## RESULTS

Muscle activation of the lateral gastrocnemius, rectus femoris, biceps femoris, gluteus medius, gluteus maximus, and tibialis anterior was monitored and recorded during a single leg squat in 5 foot positions. A percent of the maximal voluntary contraction (%MVC) was found by comparing muscle activation to the maximal voluntary contraction (MVC) value determined by taking the average MVC values of participants in the preceding study. Because assumptions were violated due to small sample size, Friedman's test of sphericity was chosen to determine statistical significance between test subjects for each muscle. our study violated assumptions which prevented us from running a parametric test. The mean and standard deviation for each foot position are listed in Tables 3.1 to 3.6, Friedman's is listed in Table 2, and Mauchley's is presented in Tables 4.1 to 4.6. Friedman's Test was run due to the study violating assumptions of a normal distribution. The post hoc tests were performed on muscles found to be significant to study the interaction between foot positions.

### Tibialis Anterior, Rectus Femoris, and Gluteus Medius

Muscle activity of the tibialis anterior, rectus femoris, and gluteus medius was not statistically significant between foot positions according to Friedman's.

### Lateral Gastrocnemius

Muscle activity of the lateral gastrocnemius was found to be statistically significant with changing foot position ($P$=.014). According to the post hoc analysis there is a significant difference in lateral gastrocnemius muscle activity between the following positions: neutral foot position and 5 degrees pronation (*P*=.050); 5 degrees supination and 10 degrees pronation (*P*=.035); 10 degrees supination and 5 degrees pronation (*P*=.020); 10 degrees supination and 10 degrees pronation (*P*=.035).

## Biceps Femoris

Muscle activity of the biceps femoris was found to be statistically significant with changing foot position (*P*=.046)  According to the post hoc analysis there is a significant difference in biceps femoris muscle activity between the following positions: neutral foot position and 5 degrees pronation (*P*=.010); neutral foot position and 10 degrees pronation (*P*=.008).

## Gluteus Maximus

Muscle activity of the gluteus maximus was found to be statistically significant with changing foot position (*P*=.000)  According to the post hoc analysis there is a significant difference in gluteus maximus muscle activity between the following positions: neutral foot position and 5 degrees supination (*P*=.021); neutral and 10 degrees supination (*P*=.023); neutral and 5 degrees pronation (*P*=.002); neutral and 10 degrees pronation (*P*=.001); 5 degrees supination and 10 degrees pronation (*P*=.003); 5 degrees pronation and 10 degrees pronation (*P*=.036).

**Table 2** Friedman's Test

| Muscle | N | Chi-square | df | *P* (significance) |
|---|---|---|---|---|
| Anterior tib | 10 | 2.240 | 4 | 0.692 |
| Lateral Gastroc | 10 | 12.560 | 4 | 0.014 |
| Rectus Femoris | 10 | 5.246 | 4 | 0.263 |
| Biceps Femoris | 10 | 9.680 | 4 | 0.046 |
| Glut Medius | 10 | 6.080 | 4 | 0.193 |
| Glut Max | 10 | 22.814 | 4 | < .001 |

Friedman's Test showing significance of muscle activity compared to %MVC for each muscle. Muscles found to be significant were the lateral gastrocnemius, biceps femoris, and gluteus maximus.

Tables 3: RM ANOVA Data – mean and std. deviation for each muscle in each position

**Table 3 - 1** Mean and Standard Deviation Values for RM ANOVA of the Tibialis Anterior in Each Foot Position.

| Muscle – position | Mean | Std. Deviation |
|---|---|---|
| Ant tib – std | 62.4500 | 11.05976 |
| Ant tib - sup5 | 55.1500 | 17.38162 |
| Ant tib – sup10 | 58.2500 | 10.97788 |
| Ant tib – pron5 | 60.3500 | 14.35752 |
| Ant tib – pron10 | 64.9200 | 19.27686 |
| The mean is expressed as %MVC. | | |

| **Table 3-2** Mean and Standard Deviation Values for RM ANOVA of the Lateral Gastrocnemius in Each Foot Position.Muscle – Position | Mean | Std. Deviation |
|---|---|---|
| Lat gastroc – std | 74.1600 | 7.02032 |
| Lat gastroc – sup5 | 76.9700 | 22.38859 |
| Lat gastroc – sup10 | 74.2200 | 13.29969 |
| Lat gastroc – pron5 | 82.9200 | 11.43871 |
| Lat gastroc – pron10 | 88.4900 | 20.72038 |
| . The mean is expressed as %MVC. | | |

| **Table 3-3** Mean and Standard Deviation Values for RM ANOVA of the Rectus Femoris in Each Foot Position.Muscle – Position | Mean | Std. Deviation |
|---|---|---|
| Rectus Femoris – std | 66.8400 | 8.87170 |
| Rectus Femoris – sup5 | 74.1200 | 31.97064 |
| Rectus Femoris – sup10 | 72.4800 | 31.55605 |
| Rectus Femoris – pron5 | 70.9500 | 28.14227 |
| Rectus Femoris – pron10 | 70.1200 | 33.81334 |
| The mean is expressed as %MVC. | | |

| **Table 3-4** Mean and Standard Deviation Values for RM ANOVA of the Biceps Femoris in Each Foot Position.Muscle – Position | Mean | Std. Deviation |
|---|---|---|
| Biceps Femoris – std | 68.4000 | 7.14967 |
| Biceps Femoris – sup5 | 76.4000 | 26.53446 |
| Biceps Femoris – sup10 | 87.6900 | 29.47422 |
| Biceps Femoris – pron5 | 81.8100 | 16.66890 |
| Biceps Femoris – pron10 | 87.1100 | 20.43828 |
| The mean is expressed as %MVC. | | |

| **Table 3-5** Mean and Standard Deviation Values for RM ANOVA of the Gluteus Medius in Each Foot Position.Muscle - position | Mean | Std. deviation |
|---|---|---|
| Glut med - std | 69.3750 | 9.00489 |
| Glut med – sup5 | 74.4050 | 23.69777 |
| Glut med – sup10 | 78.4017 | 25.43797 |
| Glut med – pron5 | 78.3583 | 19.46385 |
| Glut med – pron10 | 83.4767 | 24.96290 |
| The mean is expressed as %MVC. | | |

| **Table 3-6** Mean and Standard Deviation Values for RM ANOVA of the Gluteus Maximus in Each Foot Position. | | |
|---|---|---|
| Muscle – position | Mean | Std. Deviation |
| Glut Max – std | 68.6700 | 5.41542 |
| Glut Max – sup5 | 81.7700 | 16.31421 |
| Glut Max – sup10 | 91.9700 | 29.94955 |
| Glut Max – pron5 | 87.2100 | 16.35240 |
| Glut Max – pron10 | 101.4800 | 22.43251 |
| The mean is expressed as %MVC. | | |

Tables 4: Mauchley’s Test of Sphericity (tests of within-subjects effects)

| **Table 4-1** Mauchley’s Test of Sphericity for Anterior Tibialis. | | | | | |
|---|---|---|---|---|---|
| Muscle – ant tib | Type III Sum of Squares | df | Mean Square | F | Sig. (*P*) |
| Sphericity assumed | 566.655 | 4 | 141.664 | .827 | .517 |
| Error – sphericity assumed | 6165.513 | 36 | 171.264 | | |
| Mauchley’s Test of Sphericity demonstrating non-significance (*P*=.517) | | | | | |

| **Table 4-2** Mauchley’s Test of Sphericity for Lateral Gastrocnemius. | | | | | |
|---|---|---|---|---|---|
| Muscle – lat gastroc | Type III sum of squares | df | Mean square | F | Sig. |
| Sphericity assumed | 1552.019 | 4 | 388.005 | 3.060 | .029 |
| Error - sphericity assumed | 4564.565 | 36 | 126.793 | | |
| Mauchley’s Test of Sphericity demonstrating significance (*P*=.029) | | | | | |

| **Table 4-3** Mauchley’s Test of Sphericity for Rectus Femoris | | | | | |
|---|---|---|---|---|---|
| Muscle – rectus femoris | Type III Sum of Squares | df | Mean square | F | Sig. |
| Sphericity assumed | 299.593 | 4 | 74.898 | .335 | .853 |
| Error – sphericity assumed | 8053.567 | 36 | 223.710 | | |
| Mauchley’s Test of Sphericity demonstrating non-significance (*P*=.852) | | | | | |

| **Table 4-4** Mauchley's Test of Sphericity for Biceps Femoris**.** | | | | | |
|---|---|---|---|---|---|
| Muscle – biceps femoris | Type III Sum of Squares | df | Mean Square | F | Sig. |
| Sphericity Assumed | 2600.867 | 4 | 650.217 | 2.727 | .044 |
| Error – sphericity assumed | 8582.189 | 36 | 238.394 | | |
| Mauchley's Test of Sphericity demonstrating significance (*P*=.044) | | | | | |

| **Table 4-5** Mauchley's Test of Sphericity for Gluteus Medius. | | | | | |
|---|---|---|---|---|---|
| Muscle – glut med | Type III sum of squares | df | Mean square | F | Sig. |
| Sphericity assumed | 1063.610 | 4 | 265.903 | 2.695 | .046 |
| Error - sphericity assumed | 3551.358 | 36 | 98.649 | | |
| Mauchley's Test of Sphericity demonstrating significance (*P*=.046) | | | | | |

| **Table 4-6** Mauchley's Test of Sphericity for Gluteus Maximus. | | | | | |
|---|---|---|---|---|---|
| Muscle- glut max | Type II sum of squares | df | Mean square | F | Sig. |
| Sphericity assumed | 5947.152 | 4 | 1486.788 | 8.696 | <.001 |
| Error - sphericity assumed | 6155.176 | 36 | 170.977 | | |
| Mauchley's Test of Sphericity demonstrating significance (*P*=.001) | | | | | |

# CHAPTER V

# DISCUSSION

In general, our study demonstrates that muscle activity of the leg, thigh, and hip increases in supinated and pronated subtalar positions when compared to neutral. Percent MVC increased the most in four of the six tested muscles which includes anterior tibialis, lateral gastrocnemius, gluteus maximus and gluteus medius. This suggests that pronation increases lower extremity muscle activity the most increasing susceptibility to ACL injury. However, we are unable to confidently state whether a supinated or pronated position influences these muscles in a way that is related to ACL injuries due to high variability and large standard deviations. Nor are we able to pin point any muscles which, with problematic activation, may influence ACL injuries.

## Study Limitations

Our study has a number of limitations that devalue our findings and limit the ability to generalize results to other populations. The sample size was small. Test subjects were selected based on convenience and were limited to college age, healthy people. Results of the statistical analysis are limited due to high variability and large standard deviations.

The study is also limited due to the lack of similarity to real life occurrence of ACL injuries and the inability to confidently report participant specific %MVCs. Percent MVC was determined by comparing the muscle activity of each muscle of study participants during squatting activity to an average MVC determined through past study participants. Therefore, muscle activity during squatting for each individual was not compared to their own measured MVC. It is not possible to confidently say that the readings are specific to each participant and that comparison of values describes true results.

When examining a specific component of ACL injury causative factors, it is necessary to attempt to control other variables influencing ACL injury. Our study looked solely at foot position during a squat, a specific component believed to influence ACL injury. Squatting was performed in a much more controlled manner than during typical motion associated with ACL injuries. It may be difficult to draw a precise comparison from our study to dynamic ACL injuries due to the lack of similarity between study methods and real life occurrence of ACL injury.

# CHAPTER VI

## CONCLUSION

Lower extremity muscle activity during a single leg squat on the dominant leg is influenced by foot position. However, our study was inconclusive in determining which subtalar position, supination or pronation, influences muscle activity in a way that places one at risk for an ACL injury. Further research in this area is needed.

## REFERENCES


1. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. Am J Sports Med 1995;23:294-701.

2. Hewett TE, Lindenfeld TN, Riccobene JV, Noyes FR. The effect of neuromuscular training on the incidence of knee injury in female athletes. A prospective study. Am J Sports Med. 1999;27:699-706.

3. de Loes M, Dahlstedt LJ, Thomee R. A 7-year study on risks and costs of knee injuries in male and female youth participants in 12 sports. Scand J Med Sci Sports. 2000;10:90-97.

4. Feedman KB, Glasgow MT, Glasgow SG, Bernstein J. Anterior cruciate ligament injury and reconstruction among university students. Clin Orthop Rel Res. 1998;356:208-212.

5. Hewett TE, Ford KR, Hoogenboom BJ, Myer GD. Understanding and preventing ACL injuries: current biomechanical and epidemiologic considerations- update 2010. North Am J Sports Phys Ther. 2010;5:234-251.

6. 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. 1995; 23(6):694-701.

7. Malone TR, Hardaker WT, Garrett WE, Feagin JA, Bassett FH. Relationship of gender to anterior cruciate ligament injuries in intercollegiate basketball players. J South Orthop Assoc. 1993;2(1):36-39.

8. Myklebust G, Maehlum S, Hom 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.

9. Mihata LC, Buetler AI, Boden BP. Comparing the incidence of anterior cruciate ligament injury in collegiate lacrosse, soccer, and basketball players: implications for anterior cruiate ligament mechanism and prevention. Am J Sports Med. 2006;34(6):899-904.

10. Ostenberg A, Roos H. Injury risk factors in female European football: a prospective study of 123 players during one season. Scand J Med Sci Sports. 2000;10(5):279-285.

11. Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of kene joint motion patterns between men and women in selected athletic tasks. Clin Biomech (Bristol, Avon). 2001;16(5):438-445.

12. Griffin LY, Agel J, Albohm MJ, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. J Am Acad Orthop Surg. 2000;8:141-150.

13. Arendt E, Agel J, Dick R. Anterior cruciate ligament injury patterns among collegiate men and women. J Athl Train. 1999;34:86-92.

14. Rovere GD, Adair DM. Anterior cruciate-deficient knees: a review of the literature. Am J Sports Med. 1983;11:412-419.

15. Arms S, Pope MH, Johnson RJ, Fischer RA, Arvidson I, Eriksson E. The biomechanics of anterior cruciate ligament rehabilitation and reconstruction. Am J Sports Med. 1984;12:8-18.

16. Myer G, Brent JL, Ford KR, Hewett TE. Real-Time Assessment and Neuromuscular Training Feedback Techniques to Prevent Anterior Cruciate Ligament Injury in Female Athletes. J Strength Cond. 2011;33:21-35.

17. Carson D, Ford K. Sex differences in knee abduction during landing: a systematic review. Athl Train. 2011;3:373-382.

18. Myer G, Ford K, Hewett T. Rationale and clinical techniques for anterior cruciate ligament injury prevention among female athletes. J Athl Train. 2004;39:352-364.

19. Boden BP, Dean GS, Feagin JA Jr, Garrett WE Jr. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000;23:573-578.

20. Myer GD, Chu DA, Brent JL, Hewett TE. Trunk and hip control neuromuscular training for the prevention of knee joint injury. Clin Sports Med. 2008;27:425-448.

21. Hodges PW, Richardson CA. Contraction of the abdominal muscles associated with movement of hte lower limb. Phys Ther. 1997;77:132-142.

22. Winter D. Biomechanical and Motor Control of Human Movement. New York, NY: John Wiley & Sons Inc; 2005.

23. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. The effects of core proprioception on knee injury: A prospective biomechanical-epidemiological study. Am J Sports Med. 2007;35:368-373.

24. Zazulak BT, Ponce PL, Straub SJ, Medvecky MJ, Avedisian L, Hewett TE. Gender comparison of hip muscle activity during single-leg landing. J Orthop Sports Phys. 2005;35:292-299.

25. Ford KR, Myer GD, Smith RL, Vianello RM, Seiwert SL, Hewett TE. A comparison of dynamic coronal plane excursion between matched male and female athletes when performing single leg landings. Clin Biomech. 2006;21:33-40.

26. Anderson F, Pandy M. Individual muscle contributions to support in normal walking. Gait Posture. 2003;17:159-169.

27. Houck JR, Neville CG, Tome J, Flemister AS. Ankle and foot kinematics associated with stage II PTD during stance. Foot Ankle Int. 2009;30:530-539.

28. Kulig K, Popovich JM, Noceti-Dewit LM, Reischl SF, Kim D. Women with posterior tibial tendon dysfunction have diminished ankle and hip muscle performance. J Orthop Sports Phys Ther. 2011;41:687-694.

29. Chmielewski T, Rudolph K, Snyder-Mackler L. Development of dynamic knee stability after acute ACL injury. J Electrophyogr Kinesiol 2002;12:267-274.

30. Myer G, Ford K, Hewett T. Rationale and clinical techniques for anterior cruciate ligament injury prevention among female athletes. J Athl Train. 2004;39:352-364.

31. Ford KR, Myer GD, Schmitt LC, Van den Bogert AJ, Hewett TE. Effect of drop height on lower extremity biomechanical measures in female athletes. Med Sci Sports Exerc. 2008;40:S80.

32. Hewett TE, Stroupe AL, Nance TA, Noyes FR. Plyometric training in female athletes. Decreased impact forces and increased hamstring torques. Am J Sports Med. 1996;24:765-773.

33. Ford K, Myer G, Hewett T. Valgus knee motion during landing in high school female and male basketball players. Med Sci Sports Exerc. 2003;35:1745-1750.

34. Boden BP, Dean GS, Feagin JA Jr, Garrett WE Jr. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000;23:573-578.

35. Sell TC, Ferris CM, Abt JP, et al. Predictors of proximal tibia anterior shear force during a vertical stop-jump. J Orthop Res. 2007;25:1589-1597.

36. Lloyd D and Buchanan T. Strategies of muscular support of varus and valgus isometric loads at the human knee. J Biomech. 2001;34:1257-1267.

37. Withrow T, Huston L, Wojtys E, Ashton-Miller J. Effect of varying hamstring tension on anterior cruciate ligament strain during in vitro impulsive knee flexion and compression landing. J Bone Joint Surg Am. 2008; 90;815-823.

38. van Eijden TM, de Boer W, Wejs WA. The orientation of the distal part of the quadriceps femoris muscle as a function of the knee flexion-extension angle. J Biomech. 1985;18:803-809.

39. Podraza JT, White SC. Effect of knee flexion angle on ground reaction forces, knee moments and muscle co-contraction during an impact-like deceleration landing: implications for the non-contact mechanism of ACL injury. The Knee 2010;17:291-295.

40. Colby S, Francisco A, Yu B, Kirkendall D, Finch M, Garrett W Jr. Electromyographic and kinematic analysis of cutting maneuvers. Implications for anterior cruciate ligament injury. Am J Sports Med. 2000;28:234-240.

41. Elias J, Faust A, Chu Y-H, Chao E, Cosgarea A. The soleus muscle acts as an agonist for the anterior cruciate ligament. Am J Sports Med 2003;31(2)241-246.

42. Nicolas JA. Injuries to knee ligaments. JAMA 1970;212:2236-2239.

43. Uhorchak JM, Scotville CR, Williams GN, Arciero RA, St Pierre P, Taylor DC. Risk factors associated with noncontact injury to the anterior cruciate ligament: a prospective four-year evaluation of 859 West Point cadets. Am J Sports Med. 2003;31:831-842.

44. Moretz JA, Walters R, Smith L. Flexibility as a predictor of knee injuries in college football players. Phys Sportsmed. 1982;10:93-97.

45. Godshall RW. The predictability of athletic injuries: an eight-year study. J Sports Med. 1975;3:50-54.

46. Dedrick GS, Sizer PS, Merkle JN, et al. Effect of sex hormones on neuromuscular control patterns during landing. J Electromyogr Kinesiol. 2008;18:68-78.

47. Lee C, Liu X, Smith C, et al. The combined regulation of estrogen and cyclic tension on fibroblast biosynthesis derived from anterior cruciate ligament. Matrix Biol. 2004;23:323-329.

48. Boden B, Dean G, Feagin J, Garrett W. Mechanisms of anterior cruciate ligament injury. Orthopedics 2000;23(6):573-578.

49. Beynnon B, Fleming B. Anterior cruciate ligament strain in-vivo: A review of previous work. J Biomech. 1998;31:519-525.

50. Li G, Rudy TW, Sakane M, et al. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. J Biomech. 1999;32:395-400.

51. Ford KR, Myer GD,Hewett TE. Valgus knee motion during landing in high school female and male basketball players. Medicine and Science in Sports and Exercise. 2003;35:1745-1750.

52. Chappell JD, Creighton RA, Giuliani C, Yu B, Garrett WE. Kinematics and electromyography of landing preparation in vertical stop-jump: risks for noncontact anterior cruciate ligament injury. Am J Sports Med. 2007;35:235-241.

53. Hewett TE, Myer GD, Ford KR,  et al. Biomechanical measure 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.

54. Boden B, Dean G, Feagin J, Garrett W.  Mechanisms of anterior cruciate ligament injury. Orthopedics 2000;23(6):573-578.

55. Fukuda Y, Woo SL, Loh JC, et al.  A quantitative analysis of valgus torque on the ACL; a human cadaveric study.   J Orthop Res.  2003;21:1107-1112.

56. Buenen G, Malina R.  Growth and physical performance relative to the timing of the adolescent spurt.  Exerc Sport Sci Rev.  1988;16:503-540.

57. Khamis S, Yizhar Z.  Effect of feet hyperpronation on pelvic alignment in a standing position.  Gait Posture.  2007;25:127-134.

58. Beckett, M.E., Massie, D.L., Browers, K.D., Stoll, D.A.  Incidence of hyperpronation in the ACL injured knee: a clinical perspective.  Journal of Athletic Training 1992;27:58-62.

59. Halbach J.  Pronated foot disorders.  J Athl Train.  1981;16:53-55.

60. Lutter L.  Foot-related knee problems in the long distance runner.  Foot Ankle.  1980;1:112-116.

61. DeLarcerda F.  The relationship of foot pronation, foot position, and electromyography of the anterior tibialis muscle in three subjects with different histories of shinsplints.  J Orthop Sports Phys Ther.  1980;2:60-64.

62. Bellchamber TL, van den Bogert AJ.  Contributions of proximal and distal movements of axial tibial rotation during walking and running.  J Biomech.  200;33:1397-1403.

63. Hewett TE, Myer CG, Ford KR, et al.  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.

64. Loudon JK, Jenkins W, Loudon KL.  The relationship between static posture and ACL injury in female athletes.  J Orthop Sports Phys Ther.  1996;24:91-97.

65. Joseph M, Tiberio D, Baird J, et al. Knee valgus during drop jumps in national collegiate athletic association division I female athletes: the effect of a medial post. Am J of Sports Med. 2008;36:285-289.

66. Allen M, Glasoe W. Metrecom measurement of navicular drop in subjects with anterior cruciate ligament injury. J Athl Train. 2000;35:403-406.

67. Kulig K, Reischl SF, Pomrantz AB, et al. Nonsurgical management of posterior tibial tendon dysfunction with orthoses and resistive exercise: a randomized controlled trial. Phys Ther. 2009;89:26-37.

68. Ringleb SI, Kavros SJ, Kotajarvi BR, Hansen DK, Kitaoka HB, Kaufman KR. Changes in gait associated with acute stage II posterior tibial tendon dysfunction. Gait Posture. 2007;25:555-564.

69. Moore TM, Harvey JP Jr. Roentgenographic measurement of tibial plateau depression due to fracture. J Bone Joint Surg Am. 1974;56:155-160.

70. Terauchi M, Hatayama K, Yanagisawa S, Saito K, Takagishi K. Sagittal alignment of the knee and its relationship to noncontact anterior cruciate ligament injuries. The American Journal of Sports Medicine 2011 1-5.

71. Myer GD, Ford KR, Hewett TE. Rationale and clinical techniques for anterior cruciate ligament injury prevention among female athletes. J Athl Train. 2004;39:352-364.

72. Shultz SJ, Nguyen A-D, Levine BJ. The relationship between lower extremity alignment characteristics and anterior knee joint laxity. Sports Health. 2009;1(1):54-60.

73. Criswell E. Cram’s Introduction to Surface Electromyography. 2nd ed. Sudbury MA: Jones and Bartlett Publishers; 2011.