THE EFFECTS OF GENDER ON THE BIOMECHANICS OF THE HIP DURING ATHLETIC MANEUVERS

by

Mikaela Boham

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Mikaela Boham

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Ronald Pfeiffer, Ed.D.	Co-Chair, Supervisory Committee
Jonathan Brendefur, Ph.D.	Co-Chair, Supervisory Committee
John McChesney, Ph.D.	Member, Supervisory Committee
Michelle Sabick, Ph.D.	Member, Supervisory Committee
Lynda Ransdell, Ph.D.	Member, Supervisory Committee

The final reading approval of the dissertation was granted by Ronald Pfeiffer, Ed.D. and Jonathan Brendefur, Ph.D., Co-Chairs of the Supervisory Committee. The dissertation was approved for the Graduate College by John R. Pelton, Ph.D., Dean of the Graduate College.

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By Mikaela Boham

ABSTRACT

Context: Females are two to eight times more likely to sustain an ACL injury than males participating in the same sport. The primary mechanism reported for noncontact ACL injury involves landing from a jump, unanticipated change of direction, and/or deceleration activities.

Objective: The purpose of this study was to determine if adolescent female athletes perform athletic activities with decreased hip and knee flexion angles, and decreased EMG activity of the gluteus medius relative to their male counterparts.

Design: Cohort study from local club basketball teams.

Setting: University Laboratory.

Participants: Ten healthy adolescent basketball athletes (5 females, 5 males).

Interventions: Each participant was instructed to jump over a barrier, land with each foot on a floor-mounted force plate, and cut in a specific direction. Participants made a side cut either to the right or left, or stepped forward into a straight run. Each subject performed fifteen (15) randomized jump, land, and unanticipated cutting maneuvers.

Main outcome measures: The peak electromyography (EMG) and ground reaction force (GRF) [normalized with body weight] data were analyzed during the landing for the three cutting directions. Kinematic variables include joint angles for the ankle, knee and hip at landing and push off.

Analysis: Independent samples t-tests examined differences between the genders for dependent variables.

Results: No differences were noted for the left or right EMG amplitudes or muscle onsets. The joint angle in the left ankle ($p = 0.019$) during peak knee flexion of the left cut demonstrated the females performed tasks with greater dorsiflexion angles than males. However, during the peak GRF of the center cut in the right ankle ($p = 0.012$) males had greater dorsiflexion. The male participants sustained greater anterior forces in the left leg during the peak knee flexion angle ($p = 0.022$) and push off ($p = 0.040$) during the left cut. The male participants sustained lateral forces and female participants sustained medial forces ($p = 0.010$) during the center cut. The female participants sustained greater anterior forces in the right leg than the males ($p = 0.041$) during the peak knee flexion angles, and that females sustained anterior forces, while the male's sustained posterior forces ($p = 0.009$) in the right leg during peak GRF. The male participants sustained greater medial forces during the peak knee flexion angles ($p = 0.031$) compared to the female participants.

Clinical relevance: This study may advance our understanding of potential forces and muscle activation strategies about the ankle, knee, and hip during sport specific activities as our findings suggest women might sustain different forces during landing and cutting.

Even though we did not find statistical differences in the muscle activation strategies when comparing gender, further analysis could reveal muscular imbalances or muscle training issues between the genders. The females in this population were athletically trained and participated in training outside of their sport, which could decrease the gender effects seen in other studies. Additionally, this study could provide support for the screening of hip strength during the pre-participation physical examination and the education and creation of targeted exercise intervention programs designed to reduce the risk of non-contact ACL injuries.

Keywords: anterior cruciate ligament; kinematics; kinetics; knee; basketball athletes; gender-differences; hip

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CHAPTER 1

Introduction

Previous research has attempted to determine the impact of gender on an athlete's ability to perform athletic maneuvers with some promising results. When evaluating injuries overall, male and female athletes are at similar risk for injury. However, when evaluating the risk of injury for the lower extremity, researchers have discovered an apparent gender disparity in the incidence of anterior cruciate ligament (ACL) injuries. Female athletes are 2 to 8 times more likely to injure their anterior cruciate ligament (ACL) than their male counterparts (Anderson et al., 2001; Decker et al., 2003; Huston & Wojtys, 1996; Hutchinson & Ireland, 1995; Junge & Dvorak, 2004; McLean et al., 2003; Moeller & Lamb, 1997; Piasecki et al., 2003; Pollard et al., 2004; Powell & Barber-Foss, 2000; Rozzi et al., 1999; Slauterbeck et al., 2002; Toth & Cordasco, 2001; Wojtys et al., 2003; Wojtys & Huston, 1994). In particular, females are far more likely to sustain a noncontact ACL injury during sports participation requiring large amounts of acceleration, deceleration, jumping, landing and/or changes of direction (Decker et al., 2003; McLean et al., 2003; Moeller & Lamb, 1997; Slauterbeck et al., 2002; Toth & Cordasco, 2001). Possible risk factors associated with noncontact ACL injuries include environmental, hormonal, anatomical and neuromuscular factors (Anderson et al., 2001; Moeller & Lamb, 1997). Gender differences have been a primary focus for many researchers, but other authors postulate the increased risk of injury is more dependent on sport specific activities rather than gender (Cowley et al., 2006).

Studies have suggested females perform athletic activity utilizing less trunk flexion (DiStefano et al., 2005; Decker et al., 2003; McLean et al., 2004b; Salci et al., 2004; Yu et al., 2006), less hip flexion (Decker et al., 2003; DiStefano et al., 2005; Ford et al., 2005; Jackson et al., 2008; Kernozek, et al., 2005; Kulas et al., 2008; McLean et al., 2004a; McLean et al., 2004b; Salci et al., 2004; Wikstrom, 2004; Yu et al., 2006), greater hip adduction (Jackson et al., 2008; Jacobs et al., 2007, Pollard et al., 2004), greater hip internal rotation (Jackson et al., 2008; Pollard et al., 2004), greater knee abduction (Barber-Westin et al., 2005; Jackson et al., 2008; Lephart et al., 2004, Pollard et al., 2004), and lower knee flexion angles (Decker et al., 2003; DiStefano et al., 2005; McLean et al., 2004a; Salci et al., 2004; Sell et al., 2006; Wikstrom et al., 2004; Yu et al., 2006) than their male counterparts. Even with all of the research floating around, researchers have failed to determine the exact mechanism of injury to predict when or how injuries occur and we have yet to develop specific training protocols to prevent these injuries from occurring. The high rate of injury, especially to the female athlete, has been an area of concern for many allied health professions, athletes, and parents.

Statement of the Problem

An injury to the anterior cruciate ligament (ACL) is severely debilitating (Malone, Hardaker, Garrett, Feagin & Bassett, 1993). ACL injuries are common in athletes as it is estimated over 100,000 ACL injuries occur annually in the United States alone (Arendt & Dick, 1995; Colby et al., 2000; Koon & Bassett, 2004; Toth & Cordasco, 2001). The annual incidence rate for the general population (non-athletic population) is 1 in 3,000, which is relatively low (Chappell et al., 2005). The ACL injury rates for athletic populations, however, are much higher. Many researchers would agree American football continues to sustain the highest rate of ACL knee injuries per participant with an estimated 2.04 injuries per 1000 athlete-exposures (Baker, 1998). An athlete-exposure is defined as any athletic participation including games, training and/or practices in which a participant could possibly sustain an injury. When previous research has examined women's sports, gymnastics ACL injury rates are listed as the highest at 1.85 per 1000 athlete-exposures (Baker, 1998; Hutchinson & Ireland, 1995). Women's soccer ACL injury rates fall close behind with reports of 1.76 injuries per 1000 athlete-exposures followed by women's basketball at 1.12 per 1000 athlete-exposures (Hutchinson & Ireland, 1995). The figures for gymnastics and women's soccer are even higher than men's wrestling at 1.68 per 1000 athlete-exposures (Baker, 1998; Hutchinson & Ireland, 1995).

An ACL injury can be described as a result of two common mechanisms, either contact or noncontact events. It has been estimated that approximately 70% of all ACL injuries are a result of a noncontact mechanism, indicating the injury occurred while the athlete was alone and was not being influenced by the impact of another athlete (Chappell et al., 2005; Griffin et al., 2000; Koon & Bassett, 2004). Nearly 30% of ACL injuries are described as resulting from some type of contact, either with another person or piece of sports equipment (Chappell et al., 2005; Griffin et al., 2000; Koon & Bassett, 2004). A consensus exists in the literature when describing the mechanism of injury in which the ACL is most often ruptured. When noncontact injuries are described in the literature, the most common mechanism for the injury process includes a sudden change of direction usually in conjunction with either acceleration or deceleration of the body (Besier, Lloyd, Cochrane, & Ackland, 2000; Ford, Myer, & Hewett, 2003; McLean, Su, van den Bogert, 2003). Researchers have spent much time, energy, resources and equipment in an effort to determine the exact components of movement resulting in an ACL injury; however, researchers have yet to identify methods of preventing injury in the female athlete (Baker, 1998; Besier et al., 2000; Besier, Lloyd, Ackland, & Cochrane, 2001). The risk of injury to the female athlete is extremely high and therefore it is imperative researchers continue to investigate the issue to develop prevention strategies. Since the body is connected through articulate joints and levers, it is essential that researchers not only look at the joint where the injury occurs, but also examines the joints above (hip) and below (ankle) for possible contributing factors. If the body has a dysfunction at a proximal or distal joint, these issues could very well be important factors, if not the cause, of the epidemic of ACL injuries to the female athlete. Researchers have begun to focus on these other joints; however, there is more work to do to examine this traumatic sports issue.

Purpose

The primary purpose of this study was to determine if significant discrepancies exist between the genders in electromyography (EMG) amplitude of the left and right gluteus medius during a jump, land and subsequent unanticipated cut (3 random directions). A secondary purpose was to examine if muscle onset times between the genders for the right and left gluteus medius muscles varied during the athletic maneuver. A third purpose was to examine joint angles at the hip, knee, and ankle during the landing and unanticipated cutting for gender differences. A fourth purpose was to compare the ground reaction forces sustained by the participants during the landing and push off phases of this sport-specific maneuver differed between the genders.

Research Hypothesis

1. The primary research hypothesis was that there are differences in muscle activation between male and female adolescent basketball athletes, with the females exhibiting less EMG amplitude than males in the gluteus medius.
2. The second research hypothesis was that there are differences in muscle onset timing between male and female adolescent basketball athletes, with the females exhibiting later EMG muscle onset times than males in the gluteus medius.
3. The third research hypothesis was that there are differences between the genders in hip, knee, and ankle joint angles during landing and push off for both dominant and non-dominant lower extremities for adolescent basketball athletes, with the female athletes maintaining a more erect posture in the hips (decrease in hip flexion angles, and increase in hip internal rotation angles) during landing.
4. The fourth research hypothesis was that there are differences in landing and push off kinetics for both dominant and non-dominant lower extremities in male and female adolescent basketball athletes, with females landing with more vertical force relative to body weight than their male counterparts.

Statistical Hypothesis

- **Primary Hypothesis: EMG Peak Amplitude for Gluteus Medius**
  - Maximum EMG amplitude for gluteus medius during landing
    - Population 1 – all female adolescent basketball players
    - Population 2 – all male adolescent basketball players
    - H_o: $\mu_1 > \mu_2$
    - H_a: $\mu_1 \leq \mu_2$
  - Maximum EMG amplitude for gluteus medius during push off
    - Population 1 – all female adolescent basketball players
    - Population 2 – all male adolescent basketball players
    - H_o: $\mu_1 > \mu_2$
    - H_a: $\mu_1 \leq \mu_2$

- **Second Hypothesis: EMG Muscle Onset for Gluteus Medius**
  - Maximum EMG muscle onset for gluteus medius during landing
    - Population 1 – all female adolescent basketball players
    - Population 2 – all male adolescent basketball players
    - H_o: $\mu_1 > \mu_2$
    - H_a: $\mu_1 \leq \mu_2$

- **Third Hypothesis: Kinematics (Joint Angles)**
  - Landing kinematics at the hip

  - Population 1 – all female adolescent basketball players
  - Population 2 – all male adolescent basketball players
  - H_o: $\mu_1 > \mu_2$
  - H_a: $\mu_1 \leq \mu_2$
- Push Off kinematics for Cut at the hip
  - Population 1 – all female adolescent basketball players
  - Population 2 – all male adolescent basketball players
  - H_o: $\mu_1 > \mu_2$
  - H_a: $\mu_1 \leq \mu_2$
- Landing kinematics at the knee
  - Population 1 – all female adolescent basketball players
  - Population 2 – all male adolescent basketball players
  - H_o: $\mu_1 > \mu_2$
  - H_a: $\mu_1 \leq \mu_2$
- Push Off kinematics for Cut at the knee
  - Population 1 – all female adolescent basketball players
  - Population 2 – all male adolescent basketball players
  - H_o: $\mu_1 > \mu_2$
  - H_a: $\mu_1 \leq \mu_2$
- Landing kinematics at the ankle
  - Population 1 – all female adolescent basketball players
  - Population 2 – all male adolescent basketball players
  - H_o: $\mu_1 > \mu_2$

        - $H_a: \mu_1 \leq \mu_2$
    - Push Off kinematics for Cut at the ankle
        - Population 1 – all female adolescent basketball players
        - Population 2 – all male adolescent basketball players
        - $H_o: \mu_1 > \mu_2$
        - $H_a: \mu_1 \leq \mu_2$

- **Fourth Hypothesis**: Kinetics (GRF)
    - Landing kinetics
        - Population 1 – all female adolescent basketball players
        - Population 2 – all male adolescent basketball players
        - $H_o: \mu_1 > \mu_2$
        - $H_a: \mu_1 \leq \mu_2$
    - Push Off kinetics
        - Population 1 – all female adolescent basketball players
        - Population 2 – all male adolescent basketball players
        - $H_o: \mu_1 > \mu_2$
        - $H_a: \mu_1 \leq \mu_2$

Assumptions

- It was assumed all athletes gave a maximal effort while participating in all testing protocols.

- It was assumed the maximal effort given by all athletes simulated a game-like situation as close as possible within the constraints of laboratory testing.
- It was assumed the athletes were not fatigued prior to the study from the scheduled sport seasons.
- It was assumed the training and weight lifting work out protocols employed by individual participants were not drastically different between or among the genders.
- It was assumed that the measures selected actually measured the constructs of interest relative to the research questions about sport specific athletic maneuvers.

Limitations

The subject pool of only male and female adolescent club basketball athletes limited this study. The study was further limited by the participant's sport, age, and experience level. The participants of this study were required to be club basketball players who had played for at least 1-2 years and were in an adolescent age population; therefore, it is difficult to apply these results to any other population. When examining the age of the participants, the males had a greater variability in age than did the females, which could have affected our results. For the males, there was a 51-month age difference from the youngest participant (13 years, 1 month) to the oldest participant (17 years, 4 months). For the females, the age range was much smaller with only a 38-month age difference from the youngest participant (13 years, 7 months) to the oldest participant (16 years, 9 months). Male and female adolescent youth do not typically mature at the exact same rate; therefore, these larger variations in ages for the males might have an effect of the outcome of this study.

Another potentially constraining variable was the location of the research athletic activity. The participants were required to perform tasks within a university laboratory setting and not in a basketball gymnasium where they would traditionally be practicing their sport. The laboratory does not allow participants to reach maximum acceleration during the running maneuver without running into obstacles; therefore, participants might have altered the task constraints to fit the laboratory conditions. Also the presence of unfamiliar laboratory personnel could have altered the motivation and outcome of results. The participants performed the cutting task without a basketball as well which could have changed their movement execution pattern. In addition, the participants were asked to perform these landing and cutting tasks without shoes to eliminate shoe-surface interface traction issues as a potential confounding variable. Astroturf was fitted to the force plates in order to enhance the comfort of the athlete during landing.

Due to equipment malfunctions beyond our control, the number of participants with usable data is very low (10 total participants; 5 male and 5 female). As such, these numbers could have an effect on the power of the statistical measures used in this study. In order to compare gender, we employed multiple independent sample t-tests. Typically when researchers use this many t-tests, they apply a bonferroni correction for multiple tests to correct for the possibility of encountering a Type II statistical error (β error, or a false negative, which essentially means we fail to reject the null hypothesis when the null hypothesis is in fact false). A Type II error can occur when multiple tests are run and the a priori α level is not adjusted causing the researchers to failing to observe a difference when in truth there really is one. The statistical measures were performed with an $\alpha = 0.05$ without an adjustment, as we would have most likely failed to find any significance with an adjustment (bonferroni adjustment is equal to the number of tests divided by the α level). As such, we acknowledge this as a potential limitation in the study.

With additional participants in this study, we could see some of these limitations be reduced such as the age variance being larger in the males than the females, and different or corrected statistical measures could be run to determine differences between the groups to determine if the significant values we saw in this population are actually present in a larger population or specific to this limited sample.

Delimitations

The study and results will be applicable only to male and female adolescent basketball athletes due to the limited participant pool.

Definition of Terms

- **Contact Mechanism of Injury:** injury resulting from a collision or hitting another player or object
- **Ground Reaction Forces:** the forces acting on the body resulting from contact with the ground
- **Isokinetics:** mechanical system of resistance, which controls the velocity of a joint of an extremity through its arc of motion.
- **Isometric Contractions:** muscular force production without any joint movement.
- **Kinematics:** description of the joint angles producing motion of the body

- **Kinetics:** description of the forces producing motion of the body
- **Landing Mechanics:** kinematic and kinetic forces occurring as a result of landing from a jump or pushing off to begin a cutting maneuver
- **Maximum Voluntary Isometric Contraction (MVIC):** muscle activation produced for a single muscle (or group of muscles) during a maximal isometric contraction
- **Neuromuscular Factors:** factors affecting the actions of nerves and muscles in the body to produce a movement
- **Noncontact Mechanism of Injury:** injury which occurs while jumping, landing, decelerating, or cutting without contacting or colliding with another player or object

CHAPTER 2

Review of Literature

Sports History

Title IX states: “No person in the United States shall, on the basis of sex, be excluded from participation in, be denied the benefits of, or be subjected to discrimination under any education program or activity receiving Federal financial assistance” (Title IX of the Education Amendments of 1972, P.L. 92-318, 20 U.S.C.S. section 1681 et seq. was enacted on June 23, 1972. Its language is patterned after the preexisting Title VI as referenced in Title IX by Linda Jean Carpenter & R. Vivian Acosta, 2005, p. 3). Title IX does not explicitly state women will be given the same opportunities as men to participate in sports activities; however, “the most publicized effect of Title IX relates to interscholastic and intercollegiate athletic programs” (Carpenter & Acosta, 2005, p. 84).

A school or organization may choose one of three options to comply with the requirements of Title IX: (1) the opportunities for both male and female athletes are proportional to the general populations enrollment for the institution; (2) the organization or institution demonstrates a history and persistence in perpetuating practices of augmentation to develop the interests of the students it serves; or (3) the institution can present a functioning program to accommodate the interests of the students based on gender (Carpenter & Acosta, 2005). As such, many universities, colleges and public school systems have added women’s sports teams to comply with the demands of the laws.

The numbers of women’s intercollegiate sports teams have increased with the legal changes, and associated opportunities have created a population boom since the advent of Title IX in the early 1970s. More universities sponsor women’s athletic teams in sports like soccer, and lacrosse than similar men’s teams in 2003, according to information provided by the NCAA (Mihata, Beutler, & Boden, 2006). In 2003, the NCAA sponsored 202 men’s soccer teams and 288 women’s soccer teams (NCAA Participation, 2009). During that same time frame, there were 77 sponsored women’s lacrosse teams compared to 54 men’s lacrosse teams (NCAA Participation, 2009). The sport of basketball was nearly equal across genders with 326 men’s teams and 323 women’s teams (NCAA Participation, 2009). Although sports like soccer and basketball are not typically classified as collision sports, they both are commonly characterized as aggressive and fast paced with corresponding high frequency rates of injury. Sports such as these require rapid decision-making abilities to avoid opponents and obstacles; and therefore prevent common contact injuries. In addition to the substantial opportunity for female’s involvement in athletics, the way female athletes participate in activity has changed with time as well. The level of participation and competition availability for female athletes has increased substantially and now many female athletes are participating not only in scholastic athletic activity, but also in community and club based athletic activity (Sigward & Powers, 2006).

The way sports are played has changed over the decades of participation. “Females were not permitted to run a marathon in the Olympics until 1984” (Carpenter &

Acosta, 2005). Many prominent physicians in the early 20th Century believed that women were too dainty and unaggressive to sustain major injury (Baechele & Earle, 2000). The style of play for many current female athletes closely mimics that of their male counterparts with some female teams playing in co-ed or male tournaments and leagues. Earlier in the century, women were required to wear restrictive clothing (usually full length dresses) when participating in physical activity to maintain levels of modesty, making active aggressive sports participation nearly impossible (Baechele & Earle, 2000; Hunter, Martin, & Umberger, 2003). The evolution of women's sports has quickly gone from traditional (i.e., gender acceptable) limited or non-contact sports, such as tennis, swimming, and track-and-field to more physically demanding games such as basketball, hockey, football, wrestling, and soccer (Hunter et al., 2003).

Women were once restricted from many sports, but as of recently, have gained acceptance in many of the activities. As a result of the historic increase of female's participation in sports at all levels researchers, coaches, and parents have shifted their attention to the characteristics of injuries (Agel et al., 2007b). Decreased levels of physical training were proposed in the early 1970's to explain the predisposition of the female athlete to sport injury (Besier et al., 2001). At this pivotal time in history, anatomical differences between the genders were thought to increase the risk of injury for women's participation in athletics – much higher than their male equivalent (Baechele & Earle, 2000; Besier et al., 2001; Hahn, Foldspang, & Ingemann-Hansen, 1999). More recently, researchers have suggested that the likelihood of injury is determined by the specificity of the sport rather than being biologically predetermined (Hahn et al., 1999).

Overall the injury rate for males and females is fairly similar; however, injury rates for the anterior cruciate ligament (ACL) are one of the few exceptions with female athletes encountering a disproportionately high rate of injury. The number of girls and women sustaining this particular injury is magnified to such a rate to be of particular concern (Ahmad, Clark, Heilmann, & Schoeb, 2006; Chappell, Yu, Kirkendall, & Garrett, 2002; Cowley, Ford, & Myer, 2006; de Loes, Dahlstead, & Thomee, 2000; Ford et al., 2003; Harmon & Ireland, 2000; Heidt, Sweeterman, Carlonas, Traub, & Tekulve, 2000; Hutchinson & Ireland, 1995; Piasecki, Spindler, Warren, Andrish, & Parker, 2003; Rowe, Wright, Nyland, Carborn, & Kling, 1999; Wojtys, Ashton-Miller, & Huston, 2002a; Wojtys, Huston, Boynton, et al., 2002b; Wojtys, Huston, Schock, Boylan, & Ashton-Miller, 2003; Wojtys, Wylie & Huston, 1996). It is extremely important for professionals involved in any sports participation, especially the adolescent population as their bodies are not yet fully developed, to understand the injury risk for their athletes and to have a basic understanding of the anatomical structure of the injured ligament to determine strategies to prevent injuries in the future.

Structural Anatomy of the Knee

The knee joint is an extremely complex multiaxial joint, which must absorb and transmit forces to and from the lower extremity during weight bearing activity, which allows locomotion to occur (Lephart, Abt, & Ferris, 2002a; Lephart, Ferris, Riemann, Myers & Fu, 2002b). The knee is required to simultaneously allow gross motor movement while restricting intricate rotations during ambulation (Osternig, Caster, & James, 1995; Shultz, Houglum, & Perrin, 2005a; Shultz, Sander, Kirk, et al., 2005b; Wojtys & Huston, 1994; Wojtys et al., 1996; Wojtys et al., 2002a; Wojtys et al., 2002b).

Kinematically, the knee has six degrees of freedom which means it can move in six independent directions (three rotational and three translational) (Buckwalter, Einhorn & Simon, 1999). The knee joint can function with compressive loads up to three to four times the body weight, however, because of the bony configuration of the joint, the tolerance of shear and rotational loads is not as substantial (Shultz et al., 2005a; Shultz et al., 2005b).

The knee joint is comprised of two long lever arms (the tibia and femur), which result in considerable torque during physical activity exposing the knee to possible injury (Shultz et al., 2005a; Shultz et al., 2005b). The knee joint is comprised of "four articulations: the femur and tibia, the femur and the patella, the femur and the fibula, and the tibia and fibula" (Prentice, 2009, p. 646). The tibiofemoral joint, primarily responsible for weight bearing, allows for generally sagittal plane motion (flexion and extension) by rolling, spinning, and gliding the tibia on the femoral condyles (Shultz et al., 2005a; Shultz et al., 2005b). The tibial plateau has asymmetrical condyles, which provide minimal boney structural support to the knee joint (Shultz et al., 2005b). The soft tissue structures, largely muscles, are primarily responsible for initiating static and dynamic knee joint support (Osternig et al., 1995). Along with the bony structure supporting the knee, the ligaments surrounding and running through the knee provide structure and support as well and thus must be addressed.

Cruciate Ligaments

The musculotendinous, capsuloligamentous, and meniscocapsular structures provide restraints for the knee in addition to the bony articulations (Childs, 2002; Godlfuss, Morehouse & LeVeau, 1973; Rowe et al., 1999). The knee joint is preserved by the ligamentous, capsular, and muscular support for every movement except axial loading forces, since there are relatively no bony limitations hindering movement in the transverse plane (Shultz et al., 2005a; Shultz et al., 2005b). The anterior cruciate ligament originates on the medial surface of the lateral condyle of the femur and wraps through the joint to attach on the anterior tibial condyle (Arendt & Dick, 1995). The ACL's primary purpose is to provide restraint of the anterior translation of the tibia on the femur and more specifically to resist 80%-85% of the anterior load during locomotion (Koon & Bassett, 2004; Lephart et al., 2002a; Lephart et al., 2002b). In addition, the ACL also provides restraint of rotational (internal and external rotational stress) and frontal plane motion (valgus and varus stresses) placed on the knee (Koon & Bassett, 2004).

The ACL is considered to consist of either two or three twisted bands. Prentice (2009) suggests the ACL is comprised of the anteromedial, intermediate, and posterolateral bands, which are individually twisted together to create the ACL. Shultz et al. (2005a), who concur with the three band theory, suggest the three bundles are arranged so some part of the ligament is taut throughout the entire range of motion with the ACL being tightest in terminal knee extension and most lax when the knee is in a loose packed (knee flexed) position at approximately 45° of knee flexion. Other researchers suggest the ACL is made of two bundles, the anteromedial bundle and posterolateral bundle (Lephart et al., 2002a; Lephart et al., 2002b; Moeller & Lamb, 1997). According to this theory of two twisted bands, the posterolateral band has been described as tightest when the knee is in extension, and the anteromedial band would be tightest with the knee in flexion (Moeller & Lamb, 1997). Research indicates that both cruciate ligaments (anterior and posterior) are particularly tense in positions of extreme flexion and extreme extension of the knee (Koon & Bassett, 2004; Lephart et al., 2002a; Lephart et al., 2002b). The hamstring and quadriceps muscles are integral as secondary support for the anterior cruciate ligament (Prentice, 2009). The cruciate ligaments are integral in the functional stabilization of the knee joint during ambulation and locomotion (Baker, 1998; James, Phillip, Starch, Lockhart, & Slauterbeck, 2004).

The structural integrity of the ACL is maximized under tensile, or longitudinal, loading conditions (Woo et al., 1987; Woo, Hollis, Adams, Lyon & Takai, 1991). Under non-axial loads, or shearing, the properties of the ACL are minimized (Lyon, Woo, Hollis, Marcin, & Lee, 1989). The elevation angle of the ACL is maximized as the knee progresses into terminal extension, and the anterior tibial shear force generated by the quadriceps via the patellar tendon is transferred to the ACL and increases the shear forces to which it is subjected (Woo et al., 1999). The shear force component is conversely decreased with knee flexion angles and tensile components increase reciprocally (Lyon et al., 1989). Studies have postulated increases in peak knee flexion angles are concomitantly seen during landing activities with a flexed trunk, which might result in a reduction of ACL loading by simultaneously decreasing the anterior tibial shear forces produced by the quadriceps (Blackburn & Padua, 2008).

The primary function of the posterior cruciate ligament (PCL) is to limit hyperextension of the knee and restrict translation, both anterior and posterior, of the tibia during locomotion (Prentice, 2009). The PCL is considered to be stronger than the anterior cruciate ligament as it crosses from a wide attachment on the anterior surface of the medial femoral condyle and traverses posteriorly under the ACL to attach on the posterior lateral surface of the tibia (Prentice, 2009; Shultz et al., 2005a; Shultz et al., 2005b). The PCL is much shorter in length than the ACL, and remains fairly taut throughout the range of motion of the knee (Shultz et al., 2005a; Shultz et al., 2005b) (*Figure 2.1*).

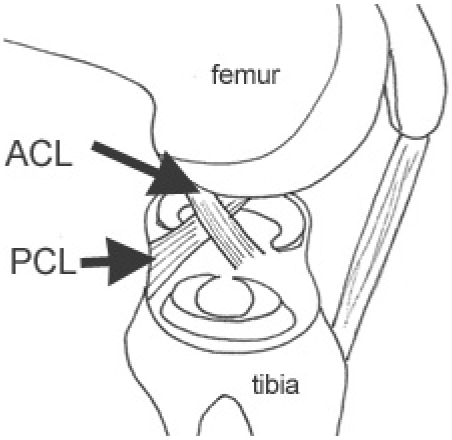


Photo: http://www.kneeguru.co.uk/assets/images/cruciates04.jpg

Figure 2.1. Anterior Cruciate Ligament (ACL) and Posterior Cruciate Ligament (PCL)

A breach of healthy cruciate ligaments, and in particular the anterior cruciate ligament, can be potentially devastating to the young athlete. Many injuries result in chronic instability and enhanced risk of meniscal or chondral damage associated with continued participation in athletics after injury (Baker, 1998). For some athletes, the injury can be career ending or result in long-term chronic issues. It is important to understand the role the ACL plays in knee movement to identify and protect the knee from risk of injury.

Functions of the ACL

The human body uses ligaments to connect the bones together to maintain stability and furnish a movement guide for the joint. Most ligaments are extremely supple and are able to withstand abundant amounts of force prior to failure and ultimate rupture. The ligaments are able to withstand traction forces because of the tensile strength of the tissues; however, they are not as responsive to rotational or torsional forces due to the formation of fiber structure. The substantial tensile strength allows ligaments to withstand considerable loads in weight bearing activity without damage or rupture. Tensile forces exceeding 500 pounds have been shown to cause a terminal rupture of knee ligaments when studied in cadaveric models (Childs, 2002). Movements such as tibial external rotation and knee valgus place the ACL in danger of rupture due to the structure of the ligament fibers (Prentice, 2009).

The ACL has five primary functions when applied to athletic activity: (1) prevention of anterior tibial translation on the femur, specifically during knee extension, (2) preventing knee hyperextension, (3) assisting with rotatory knee stability, (4) assisting in preventing valgus or varus knee laxity, (5) serving as a guide for the femur on the tibia during terminal knee extension (screw-home mechanism) (Godlfuss et al., 1973; Prentice, 2009; Rowe et al., 1999; Shultz et al., 2005a; Shultz et al., 2005b). Disruption of knee ligaments is extremely devastating and traumatic. Unfortunately, it is not uncommon for an athlete to damage other structures when an ACL injury occurs. Damage to the knee is typically a very traumatic event, even when there is no contact involved, and as such the damage is not always localized to one specific area or structure. Other structures within the knee can also sustain damage during the course of injury including the collateral ligaments.

Medial Collateral Ligament

The medial collateral ligament (MCL) is located on the medial aspect of the knee. The ligament originates on the medial epicondyle of the femur and along portions of the tibia and inserts inferior to the pes anserine along the medial tibial flares (Prentice, 2009). The primary function of the MCL is to restrict valgus and external rotation forces at the knee (Prentice, 2009). Athletic trainers, physical therapists, and physicians assess MCL injuries by asking athletes to identify a history of medial knee pain and have had a reported mechanism of injury in which a valgus force was applied to the lateral aspect of the tibiofemoral joint (Aronson, Rijke, & Ingersoll, 2008). Clinical diagnosis of MCL injury severity is based on the patient's report of point tenderness and/or swelling over the ligament and the amount of laxity at the medial joint line during manual abduction, also referred to as a valgus stress test (Aronson et al., 2008; Hillard-Sembell, Daniel, Stone, Dobson, & Fithian, 1996; Petersen & Laprell, 1999). Kennedy and colleagues (1977) state the tautest position of the MCL occurs in hyperextension (genu recurvatum). Subsequently the most laxity for the MCL occurs when the knee is in full flexion (Kennedy, Hawkins, & Willis, 1977).

Grood and associates (1981) determined the MCL decreased valgus stress on the knee during full knee extension by about 50% while the capsule contributed nearly 25%, and the cruciate ligaments together assisted by approximately 25% (Grood, Noyes, Butler, & Suntay, 1981). The MCL can be divided into three functional units as it acts as a tibiofemoral joint stabilizer: (1) the superficial medial portion which includes the anterior border of the ligament, (2) the deep medial portion, which includes the meniscofemoral and meniscotibial ligaments (middle capsular ligament), and (3) the posterior oblique fibers, which blend into the knee's posterior joint capsule (Robinson, Sanchez-Ballester, Bull, Thomas, & Amis, 2004; Slocum, Larson, & James, 1974; Warren, Marshall, & Girgis, 1974).

As the knee flexes, the MCL slides posteriorly and the anterior portion of the ligament becomes taut in nearly all degrees of joint flexion during the movement (Arms, Boyle, Johnson, & Pope, 1983; Brantigan & Voshel 1941; Meister, Michael, Moyer, Kelly & Schneck, 2000). As the knee is flexed, the posterior portion of the ligament becomes slack (Grood et al., 1981; Shapiro, Markolf, Finerman, & Mitchell, 1991). When the knee extends, the MCL moves anteriorly and pulls both the anterior and posterior portions taut (Brantigan & Voshell, 1941). Valgus movement of the tibiofemoral joint is prevented by the MCL when the knee is in terminal knee extension (Arms et al., 1983; Grood et al., 1981; Haimes, Wroble, Grood, & Noyes, 1994; Pressman & Johnson, 2003; Seering, Piziali, Nagel, & Schurman, 1980). The ligaments of the knee assist with joint range of motion in addition to bony structures and cartilage support.

Menisci

The menisci provide cushioning and shock absorption for the knee joint and preserve space between the tibial plateau and the femoral condyles. The menisci improve the weight distribution and increase contact area between the tibia and femur resulting in increased joint stabilization and augmenting the contact with the articular surface of the tibia (Shultz et al., 2005a; Shultz et al., 2005b). The medial and lateral menisci are comprised of fibrocartilage disks housed on the superior surface of the tibial condyles (Starkey & Johnson, 2006). The menisci are not bilaterally symmetrical as the lateral meniscus is nearly a complete ring, whereas the medial meniscus is more of a crescent or C-shaped structure (Prentice, 2009). Damage to the meniscus creates a potentially devastating injury, as the meniscus is largely avasucalar and thus has a limited regenerative capability. Injury to the menisci affects the stability of a joint and results in altered biomechanical function possibly predisposing the athlete to further injury (Shultz et al., 2005a; Shultz et al., 2005b). A medial meniscus injury can occur due to rupture of the medial collateral ligaments deep fiber attachment to the periphery of the meniscus or from a valgus force being placed on the lateral side of the knee (Shultz et al., 2005a; Shultz et al., 2005b). The lateral collateral ligament, on the other side of the knee, has no attachment to the lateral meniscus. The medial meniscus is more commonly torn in conjunction with acute ACL injuries and the lateral meniscus injury more commonly results from chondral defects occurring in the chronically ACL-deficient knee (Fithian, Paxton, & Goltz, 2002; Starkey & Johnson, 2006). Occasionally an athlete has the misfortune of sustaining traumatic injuries beyond the scope of an ACL rupture, including damage to the medial collateral ligament and medial menisci which is commonly referred to as an Unhappy Triad.

Unhappy Triad

Grade three ruptures of ligaments result in a complete tear of the supporting tissue. Major signs and symptoms include: complete loss of stability, minimum to moderate swelling, immediate severe pain followed by a dull ache, loss of motion as a result of effusion and guarding (Prentice, 2009). Grade three sprains of the MCL most frequently result from a direct valgus force while the foot is planted in full weight bearing activity. Nearly all MCL injuries occurring in conjunction with rotational forces result in injury to the ACL, medial meniscus and occasionally the PCL as well (Prentice, 2009). When an athlete sustains a valgus load resulting in injury to the ACL, MCL, and medial meniscus the condition is referred to as an “Unhappy Triad” or the “Unhappy Triad of O’Donohue” (Shultz et al., 2005a; Shultz et al., 2005b). Injuries to the ACL and surrounding structures frequently require surgical intervention. Athletic trainers are taught to evaluate the injuries sustained to a person; however, we sometimes forget the body is a linked chain; therefore it is vitally important to evaluate the joints above and below the knee as well.

Structural Anatomy of the Hip, Thigh and Pelvis

The hip and pelvis are among the largest, strongest and most stable joints in the body (Shultz et al., 2005a; Shultz et al., 2005b). The ligaments, joint capsule, and musculature surrounding the hip are extremely strong; however, due to the large amount of movement allowed by the ball-and-socket joint, the area suffers frequent injuries (Prentice, 2009). Injuries to the hip and thigh are less common than those occurring at the knee and ankle; however, injuries are still traumatic and could lead to the increased risk of injury from other structures above or below the hip joint (Shultz et al., 2005a; Shultz et al., 2005b). The hip sockets are composed of two innominate bones consisting of three parts each (ilium – positioned superiorly and posteriorly; ischium – located inferiorly; and pubis – forms the anterior junction between right and left bones) (Prentice, 2009;

Shultz et al., 2005a; Shultz et al., 2005b). The innominate bones of the pelvis ossify and typically fuse around 12-16 years of age (Prentice, 2009). The two bones articulate anteriorly via the symphysis pubis; and articulate posteriorly with the sacrum composed of five fused vertebrae to create the sacroiliac joints (Prentice, 2009; Shultz et al., 2005a; Shultz et al., 2005b). The acetabulum is a concave formation in the ischium in which the head of the femur articulates to form a ball-and-socket joint. The acetabulum serves as an attachment site for the acetabular labrum, a cartilage structure, which deepens the surface area available for femur-acetabulum coupling.

The hip and pelvic joints include the coxofemoral, sacroiliac, and pubic symphysis, which function as a complex unit to provide stability for the torso as well as mobility for locomotion (Shultz et al., 2005a; Shultz et al., 2005b). The hip is a multiaxial ball-and-socket joint provided by the head of the femur and the acetabulum of the pelvis (Shultz et al., 2005a; Shultz et al., 2005b). Gross movement of the hip only happens at the coxofemoral (hip) joint. The acetabulum is padded at its center by a mass of fatty tissue, ligaments and capsule forming an incomplete bony ring interrupted by a notch on the lower aspect of the socket (Prentice, 2009). The ring is completed with the attachment of the transverse ligament as it crosses the inferior notch (Prentice, 2009). “The femoral head is a sphere that fits into the acetabulum in a medial, upward, and slightly forward direction” (Prentice, 2009, p. 711).

The hip is a highly mobile structure permitting movement in nearly all planes of movement, including: flexion, extension, abduction, adduction, circumduction, and rotation (Shultz et al., 2005a; Shultz et al., 2005b). The limited motions available at the sacroiliac and pubic symphysis joints provide structural support for the hip. During locomotion, the hip typically moves in all three planes of motion: sagittal (flexion/extension), frontal (abduction/adduction), and transverse (rotational component) planes (Prentice, 2009). Weakness, muscular imbalance, and reduced flexibility are predisposing factors for injury (Starkey & Johnson, 2006). The forces placed on the hip during running have been documented to reach up to five times the body's normal weight suggesting impact loads possibly contributing to injuries of both muscle and bone (Prentice, 2009).

The iliofemoral (Y), pubofemoral, and ischiofemoral ligaments provide structural stability to the hip as well (Shultz et al., 2005a; Shultz et al., 2005b). The hip is supported by approximately twenty-two muscles, which are typically classified by the muscular actions they produce.

> The muscles are described as "three flexors (psoas, iliacus, and rectus femoris); one flexor adductor (pectineus); three extensors (biceps femoris – long head, semimembranosus, semitendinosus); one extensor outward rotator (gluteus maximus); one abductor (gluteus medius); four adductors (gracilis, adductor longus, brevis and magnus); two inward rotators (tensor fascia latae, gluteus minimus); six outward rotators (piriformis, obturator externus, obturator internus, gemelli superior and inferior, quadrates femoris); and one flexor-abductor outward rotator" (Sartorius) (Shultz et al., 2005a, p. 478).

To prevent injuries or to reduce the risk of injury to the hip, it is essential to maintain strength and flexibility in the muscles surrounding and attaching on the hip, thigh and pelvis (Prentice, 2009). Functional limitation of the adductor muscle of the hip can result in the inability to function at a normal level in sports participation (Starkey & Johnson, 2006). The adductors stabilize the hip during activities and loss in strength can create muscular imbalances, which could lead to hip and knee injuries (Starkey & Johnson, 2006).

The pelvis itself has the ability to move in three directions: anteroposterior tilting, lateral tilting, and rotation (Prentice, 2009). During movement, the hip joint acts as the center of the rotation (Golding & Golding, 2003). The hip flexors (rectus femoris & iliopsoas) create anterior tilting in the sagittal plane (Prentice, 2009). In opposition, the hamstring muscles pull on the pelvis to create a posterior tilting (Prentice, 2009). Pelvic lateral tilting is a result of hip abduction and adduction, with the hip abductors controlling lateral tilting by contracting isometrically or eccentrically (Prentice, 2009). Rotation of the pelvis occurs in the transverse plane as a result of gluteal muscle, external rotator, adductor, pectineus, and iliopsoas activation (Prentice, 2009). The movements of the hip and pelvis play an important role in injury prevention and evaluation (Prentice, 2009).

Weight-bearing activities result in considerable forces transmitted through the hip and pelvis. The large site of muscular attachment surrounding the hip, thigh, and pelvis result in a variety of movements being produced by the core and lower extremity of the body and as such the muscles in this region are susceptible to injury resulting from the dynamic power-producing contractions occurring at the hip joint (Prentice, 2009). The pelvis and hip support the abdominal muscles and form the base of support for the trunk (Prentice, 2009). The pelvis and pelvic girdle link the lower extremity to the trunk as well. The muscles surrounding the hip and pelvis play a primary role in the initiation of postural stability and locomotion. When injuries are sustained to the hip or thigh, the role of the structures providing structural support and locomotion can turn even minor injuries into debilitating ones. A detailed understanding of the way the hip moves and responds to athletic activity is extremely important. If researchers could identify postures or positions increasing the risk of injury we might be able to create training protocols to decrease the likelihood of athletes being reliant on these motions.

Point of No Return

Researchers have suggested the noncontact ACL injury occurs during a specific point in the range of motion identified as the “point of no return” or the “position of no return” (Blackburn & Padua, 2008; Hart, Garrison, Kerrigan, Palmieri-Smith, & Ingersoll, 2007; Ireland, 1999; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007). The “point of no return” is described a position consisting of hip adduction and internal rotation, knee valgus and external tibial rotation, and subtalar pronation (Blackburn & Padua, 2008; Ireland, 1999). Female athletic populations typically display simultaneous decreased hip, knee, and trunk flexion during gait and landing tasks compared with male athletes (DiStefano, Padua, Prentice, Blackburn, & Keras, 200; Decker, Torry, Wyland, Sterett, & Steadman, 2003; McLean, Huang, Su, et al., 2004a; McLean, Lipfert, & van den Bogert, 2004b; Yu et al., 2006), implicating sagittal-plane coupling of these joints could play a role in the mechanism of injury to the knee ligaments. Increased planar motion of the knee has been implicated in the ACL ligament sprain and ultimate terminal injury of the ligament (Jacobs et al., 2007).

Ford et al. (2006) and Kernozek et al. (2005) have each reported a significant increase in frontal plane motion of the female athlete’s knee when landing compared with their male counterparts (Kernozek, Torry, Van Hoff, Cowley, & Tanner, 2005). The increase in frontal plane motion could result from inferior hip abductor function, which could create a knee position similar to the point of no return (Jacobs et al., 2007). In addition, the gluteus medius (the primary hip abductor muscle) has been reported to provide support to the pelvis and hip during midstance of the gait (Anderson & Pandy, 2003). The gluteus medius muscle has also been postulated to control femoral internal rotation movement during activity. Without significant muscular strength in the gluteus medius, an athlete might not be able to functionally control hip adduction and internal rotation during athletic tasks (Hart et al., 2007). The suggested point of ACL injury has been highly associated with both hip adduction and internal rotation moments thus implying decreased muscular activity to the "point of no return" at the hip could expose the knee to injury (Hart et al., 2007; Ireland, 1999).

Injury History

Risk of Injury

The number of youth athletes participating in sports has risen dramatically over the past few decades, and so has the number of ACL injuries. Scholars have suggested children, especially girls, are now participating in athletic maneuvers commonly associated with ACL injury. ACL injury has been described as occurring during deceleration or change of directional forces, which frequently occurs while playing sports like basketball, football, and soccer, but is not limited to only these sports or activities (Bach, Chapman, & Calvert, 1983; Bach, Hull, & Patterson, 1997).

Researchers suggest the overall rate of injury in intercollegiate athletes is relatively low with one injury reported in every two games played or one injury in every five practices in which athletes participate (Hootman, Dick, & Agel, 2007). In addition, a substantial proportion of injuries are minor in nature and do not result in significant time lost to the athlete during recovery (Hootman et al., 2007). The number of ACL injuries has increased proportionally with the explosion of female participants over the past nearly four decades since the passage of Title IX (Ford, Myer, Toms, & Hewett, 2005). Data for intercollegiate athletics is collected via the Injury Surveillance System; however, no such system exists at lower levels of athletic participation including the elementary school, junior or middle school, and high school athletes. McCarroll (1994) suggested nearly 3.3% of all ACL injuries occur in the skeletally immature sports participant. The rates of injury for athletic club participants are difficult to discern due to the lack of reporting organizations.

Female adolescent athletes participating in sports which require the athlete to sustain multiple pivoting and jumping activities appear to sustain anterior cruciate ligament (ACL) injuries at rates nearly 2-8 times greater than their adolescent male counterparts (Decker et al., 2003; Anderson, Dome, Guatam, Awh, & Rennirt, 2001; Hewett, 2000; Hewett et al., 1999; Huston & Wojtys, 1996; Hutchinson & Ireland, 1995; Junge & Dvorak, 2004; McLean et al., 2003; Moeller & Lamb, 1997; Piasecki et al., 2003; Pollard, Davis, Hamill, 2004; Powell & Barber-Foss, 2000; Rozzi, Lephart, Gear & Fu, 1999; Slauterbeck et al., 2002; Swartz, Decoster, Russell & Croce, 2005; Toth & Cordasco, 2001; Wojtys et al., 2003; Wojtys et al., 2002a; Wojtys et al., 2002b; Wojtys et al., 1996). Injury estimates have suggested 70% of all reported ACL injuries occur in conjunction with some sporting activity (Colby et al., 2000; Kirkendall & Garrett, 2000).

Female athletes sustain ACL injuries more commonly than male athletes (Shea, Pfeiffer, Wang, Curtin, & Apel, 2004). Until a few decades ago, many professionals thought midsubstance tears of the ACL were not probable in children and adolescent athletes (Angel & Hall, 1989; De Lee & Curtis, 1983); however, recent studies have reported an increased incidence ratio in young athletes disputing this claim (Bales, Guettler, & Moorman, 2004; Dorizas & Stanitski, 2003). Scholars have identified several causative factors to account for the gender difference including muscle strength, quadriceps-angle (Q-Angle), ligamentous laxity, and biomechanical analysis of movement (Shea, Apel, Pfeiffer, 2003).

In 1998, Stanitski studied 70 children who presented with traumatic knee effusions and determined isolated ACL tears occurred in 47% of patients, meniscal tears were reported for 30% of injuries, a combined ACL/meniscus injury was reported 16% of the time, and finally osteochondral fractures represented 7% of the study participants (Stanitski, 1998). In 2003, Luhmann conducted a similar study and found 29% of the 44 adolescent athletes (under age 18) reported ACL injuries, while 25% of ACL injuries also coincided with a meniscal injury, and osteochondral fracture represented 4% of subjects in the study. Researchers are not entirely certain why female athletes appear to be more vulnerable; however, physical attributes might be contributing factors (Luhmann, 2003; Luhmann, Schootman, Gordon, et al., 2005).

Injuries involving the knee are some of the most common traumatic damage occurring as a result of participation in sports. Knee injuries have been suggested to account for up to 91% of season ending injuries and up to 94% of injuries requiring surgical intervention in female basketball athletes, demonstrating the devastating impact of the sports-related injury (Chandy & Grana, 1985; Ford et al., 2003). Injury rates have been calculated to determine the risk of injury during sport participation by comparing male and female injury incidence. These results suggest a female collegiate basketball player is 4.1 times more likely to tear her ACL than would a similarly trained male collegiate basketball player (Anderson et al., 2001; Baker, 1998). When comparing soccer, a female collegiate soccer athlete is at a lower but still elevated risk of injury, as they are 2.4 times more likely to sustain an ACL injury than are males playing the same sport with the exact same rules, equipment, and player contact risk (Anderson et al., 2001; Baker, 1998).

The risk of injury to the female knee is not limited to traditional sports participation. The United States Military academies have assessed data on the injury rates and suggest a similar gender discrepancy exists within military training activities such as participation in the obstacle course and intramural sports participation (Gwinn, Wilckens, McDevitt, Ross & Kao, 2000; Uhorchak et al., 2003). Midshipmen at the United State Naval Academy, Cadets at the United States Military Academy and Cadets at the United States Air Force Academy are all required to participate in daily physical training activities, Department of Physical Education classes, advanced close-quarters combat skills training, and physical assessments including the obstacle course and other similar activities, as well as the traditional academic duties required of college students (Uhorchak et al., 2003). Students are required to participate in intramural, club or varsity sports during at least six of their eight semesters at the military academies (Gwinn et al., 2000; Uhorchak et al., 2003).

Military training closely resembles traditional sports activity and requires participants to engage in activities including starting and stopping, changes in direction, events requiring the person to react to outside and environmental stimuli, and perform in an efficient, fast, and effective manner to complete some physical task. Gwinn and colleagues (2000) found the relative incidence of ACL injury in women was 3.067 compared with 0.315 for men when participating in traditional military training activities (Gwinn et al., 2000). This suggests women are nearly 10 times more likely to injure their knee than are their male classmates when participating in these types of activities. Similar results were found at the United States Military Academy (USMA). Uhorchak and colleagues (2003) followed 895 cadets throughout their four-year tenure at the USMA to monitor the relative risk of ACL injury and found that a total of 29 complete ACL tears (21 in men, 8 in women) were sustained by the cadets during the four year educational cycle (Uhorchak et al., 2003). Less than 20% (5 of the 29) of the injuries were directly linked to a contact event, the remaining 80% of injuries (24 total injuries, 16 in men and 8 in women) were reported to occur from a noncontact injury mechanism (Uhorchak et al., 2003).

Women participating on intercollegiate athletic teams had a fourfold increased risk of injury compared with their male classmates (Gwinn et al., 2000). For the West Point Class of 1999 at the USMA, the overall risk of injury to the ACL was listed at 3.3%, with the incidence of noncontact ACL tears at 2.8% (6.6% in women and 2.1% in men) resulting in an injury ratio of approximately 3:1 based on gender. Of the noncontact injuries, 14 (approximately 60%) of the injuries occurred during participation in intramural sports (Uhorchak et al., 2003). Gwinn et al. (2000) reported similar statistics for female athletes and found that the relative risk of injury for female participants (playing on an all female team) in intramural soccer was up to nine times higher than that for males. When the researchers examined the coed intramural sports, the women's relative risk of injury dropped to only 1.40 times higher risk of injury than competing males (Gwinn et al., 2000). The researchers postulate women could be altering their level of aggressiveness and style of play based on the size and gender of their opponent, thus increasing their risk of injury when playing against a similarly matched opponent (Gwinn et al., 2000). The risk of injury to the knee is extremely important for coaches, medical personnel and athletes to identify in order to attempt to create prevention strategies and programs to deter injuries from occurring in the first place. In addition, trying to identify the primary mechanism of injury could allow for rule, equipment, or training regiment changes to reduce the risk of injury.

Mechanism of Injury

Anterior cruciate ligament (ACL) injuries can occur as a result of several scenarios in which the athlete either: 1) sustains contact from another individual or something in the environment, or 2) the athlete is only in contact with himself or herself when the injury occurs (Wojtys et al., 2003). The sport of basketball requires ten (10) players, five from each team, to share and to consecutively occupy space while attempting to avoid opponents and ultimately score baskets. An athlete is exposed to direct contact and noncontact forces multiple times in a game or practice setting and therefore is correspondingly exposed to potential injury. Approximately 70% of adult ACL injuries are a result of noncontact mechanism during sports participation (Besier et al., 2000; Chappell et al., 2005; Kirkendall & Garrett, 2000; Wojtys & Huston, 1994; Wojtys et al., 2003; Wojtys et al., 2002a; Wojtys et al., 2002b; Wojtys et al., 1996).

The ACL injury mechanism was studied by Boden and colleagues (2000a) for 100 skeletally mature athletes participating in activities other than skiing and determined 72%

of athletes sustained injury through a noncontact injury mechanism and the remaining 28% were the result of some form of direct contact mechanism (Boden, Dean, Feagin, & Garrett, 2000a). Arendt & Dick (1995) and Arendt, Agel, & Dick (1993) have both reported nearly all ACL tears in female collegiate athletes are reported to occur as a result of noncontact injury mechanism (Arendt, Agel, & Dick, 1993; Arendt & Dick, 1995). In addition, a majority of the injuries female athletes incurred occurred during games, not practice, with an increase in incidence as the sports season drew to a close.

Sports requiring cutting, pivoting, acceleration, landing from a jump, or sudden stops and/or changes in direction are commonly associated with ACL injuries in sports (Ahmad et al., 2006; Baker, 1998; Chappell et al., 2002; Ford et al., 2003; Hewett, 2000; Hewett et al., 1999; McLean et al., 2003; Moeller & Lamb, 1997; Wikstrom, Powers & Tillman, 2004; Wojtys et al., 2002a; Wojtys et al., 2002b). ACL injury mechanisms are commonly described as a sudden deceleration, abrupt change in direction, hyperextension, or a collision while a valgus or varus stress is applied to the knee resulting in a shear force being placed on the knee ligaments (Baker, 1998; Besier et al., 2000; Chappell et al., 2002; Childs, 2002; Decker et al., 2003; James et al., 2004; McLean et al., 2003; Moeller & Lamb, 1997; Slauterbeck et al., 2002; Toth & Cordasco, 2001). Another common ACL injury mechanisms results from an athlete having a planted foot on the ground in conjunction with a rotational component, landing from a jump, or a change in direction (Besier et al., 2000; Ford et al., 2005). In addition, several researchers have described the knee being near full extension (Kirkendall & Garrett, 2000; Koon & Bassett, 2004).

Grade three anterior cruciate ligament (ACL) sprains occur frequently as a result of noncontact forces associated with a planted foot, cutting, and the external rotation of the lower leg (Starkey & Johnson, 2006). When the previously described unstable knee position is combined with a valgus force during the cutting motion, the ligaments are subjected to tensile forces frequently exceeding the tissue strength and ultimately resulting in damage or rupture (Baker, 1998). Depending on the sport involved, ACL injuries have also been reported to occur with a noncontact mechanism that involves landing or planting the foot with the knee extended (Arendt et al., 1993; Arendt & Dick, 1995; Baker, 1998). All of the above mechanisms of injury are common in sports such as basketball, soccer, gymnastics, and volleyball (Starkey & Johnson, 2006).

In one of the few studies utilizing a younger population, Powell and Barber-Foss (1999) created a cohort observational study of varsity high school athletes and found rebounding the basketball was the primary cause of injury to the female basketball player. Cowley et al. (2006) concurred with the findings of Powell and Barber-Foss (1999), and reported female high school basketball athletes were more likely to injure their while jumping or landing (60%) than athletes participating in soccer (25%). Rebounding a basketball frequently requires an athlete to jump in the air to meet the ball coming off of the backboard and/or rim and land which supports the jumping and landing injury mechanism reported by Cowley et al. (2006) and others (Ahmad et al., 2006; Baker, 1998; Chappell et al., 2002; Hewett, 2000; Hewett et al., 1999; McLean et al., 2003; Wikstrom et al., 2004; Wojtys et al., 2002a; Wojtys et al., 2002b). Cowley et al. (2006) reported the most frequent ACL injury mechanism in soccer appears to occur during a cutting maneuver rather than during a landing, as is the case for basketball athletes (Cowley et al., 2006). These injury mechanisms could suggest a sport specific component to ACL injury risk.

Athletes are required to perform both anticipated and unanticipated maneuvers during regular sports participation. Most athletic movements during practice or game situations occur as a reaction to sudden external stimulus, such as avoiding another player, following the movements of a ball, or avoiding sports equipment and environmental factors (i.e., bleachers, chairs on the sidelines, etc.) (Besier et al., 2001). Sports such as basketball, soccer, and football all require large amounts of jumping, landing, rapid acceleration and deceleration during participation. A good majority of athletes complete these high demand, repetitive activities without injury; however, the risk factors associated with the mechanism of ACL injury needs further research so more athletes do not have to participate with the fear of injury (Cowley et al., 2006).

Young athletes sustain the highest overall injuries during participation in soccer (21%) and basketball (20%) (Cowley et al., 2006). For collegiate athletes, the National Collegiate Athletic Association (NCAA) injury data demonstrates female athletes injured their ACL more frequently than their male counterparts as well (Moeller & Lamb, 1997). The NCAA requires certified athletic trainers report athlete injuries each year via the Injury Surveillance System and a study from 1990 to 1993 injury surveys have indicated approximately 15% of NCAA member institutions reported an average knee injury rate of approximately 1 per 1000 athlete-exposures (1 athlete-exposure meaning participation in either a game or practice) or greater than 1 injury for every 10 female athletes (Arendt, Agel, & Dick, 1993). Research suggests the ACL injury rate for female basketball players is two to five times higher for female athletes than male basketball participants (Anderson et al., 2001; Decker et al., 2003; Hewett et al., 1999; McLean et al., 2003). Similar research suggests female soccer players are two to eight times more likely to sustain ACL rupture while playing soccer than are their male counterparts (Anderson et al., 2001; Rozzi et al., 1999). Decker et al. (2003) completed a study and suggested ACL injuries in soccer and basketball were most commonly noncontact injuries resulting from a deceleration type movement such as landing from a jump. The study found 30 of the 72 (approximately 41%) injury mechanisms were reported as occurring just after a deceleration and jumping task.

Research has provided coaches, and athletic trainers with some vital information about the process of the ACL injury including mechanism and player-contact levels. It is therefore, imperative that coaches and athletic trainers now focus on the availability to train younger athletes how to perform tasks without putting themselves at an increased risk of injury.

Youth Participation in Sports

Sports participation has increased dramatically over the last four decades. Both male and female athletes are entering formalized athletic participation at very young age, and new opportunities for sports participation are increasing rapidly. Across the United States of America, over 25 million high school students participate in athletic activities annually and participation rates are continuing to increase (Adirim & Cheng, 2003; Ingram, Fields, Yard, & Comstock, 2008). Another 30 million children are estimated to participate in organized athletic programs outside of the public school organizations (Adirim & Cheng, 2003; Arendt & Dick, 1995). Some reports estimate more than half of all children from age 5 to 18 are participants of some form of organized sports program (Agel, Arendt, & Bershadsky, 2005). High school sports encourage much needed physical activity, the creation of teamwork, and participation in sportsmanship among other positive attributes, however, nearly all physical activity carries some risk of injury to the participant (Ingram et al., 2008). Recently, more and more athletes have entered the sporting arena and are participating at higher levels thereby exposing themselves to greater possibility of acute and chronic injuries. Unfortunately, of the young athletes who are participating in club, school, or recreational sports, nearly a third will sustain injuries requiring medical treatment, with injuries occurring at the ankle and knee listed as the most common culprits (Adirim & Cheng, 2003; Jones & Knapik, 1999; Moti & Micheli, 2003; Radelet, Lephart, Rubinstein, & Myers, 2002).

Over the past few decades, multiple studies have reported female athletes in both adolescent and adult populations have an increased risk (4 -8 times higher risk) of serious knee ligament injury while competing in the same sports as their male counterparts (Anderson, Messner, & Green, 1964; Arendt & Dick, 1995; Cowling & Steele, 2001; Gwinn et al., 2000; Huston & Wojtys, 1996; Lindenfeld, Schmitt, Hendy, Mangine, & Noyes, 1994; Noyes, Barber & Mangine, 1991; Noyes, Barber-Westin, Fleckenstein, Walsh & West, 2005). Malone and colleagues (1993) also suggested females were up to eight times more likely to sustain a rupture of the ACL during basketball participation, with the most common injury mechanism identified as a noncontact incident. Researchers have yet to determine if similar differences exist in knee ligament injury rates for younger, skeletally immature children. Overall, it is estimated nearly 300,000 knee injuries in male athletes and over 160,000 knee injuries in female athletes nationally in the traditional 9 sports of interest at high schools: 5 boys' sports (football, soccer, basketball, baseball, and wrestling) and 4 girls' sports (soccer, volleyball, basketball, and softball) (Ingram et al., 2008).

Adolescent and prepubescent athletes most commonly injure the lower extremity (ankle and knee) followed by the hand, wrist, elbow, shin and calf, and head and neck (Radelet et al., 2002). Acute injuries are common with young athletes sustaining contusions and strains most frequently. Athletes who are in early adolescence sustain apophysitis or strains at the apophyses with the most common injuries including Osgood-Schlatter disease (knee), Sever's disease (heel), and Little League Elbow (elbow) (Adirim & Cheng, 2003).

According to sources, children and adults could have physical and physiological differences which could leave younger athletes exposed to injury including: larger surface area to mass ratios than adults; disproportionately larger heads; improperly fitting protective equipment; growing cartilage vulnerable to stress; and decreased ability to perform complex motor skills required for successful participation in many sports (Adirim & Cheng, 2003). According to Junge and Dvorak (2004), many children do not have the ability to master complex motor skills until later in childhood development, typically around 10-12 years of age (Jones & Knapik, 1999). In addition, a temporary decline in coordination and balance occurs to a majority of pubescent athletes. It has been postulated that pre-pubescent athletes generate lower speeds, have less mass, and declined strength capability and are thus less likely to sustain injury due to acute blunt trauma (Moti & Micheli, 2003).

Sports-related injuries account for an estimated 2.5 million emergency room visits annually for adolescent athletes (Moti & Micheli, 2003). Some injuries are relatively minor including contusions, sprains, and strains but the more serious injuries typically require immediate medical attention including fractures, dislocations, and torn ligaments (de Loes et al., 2000). Some researchers have suggested an adolescent athlete is more susceptible than a prepubescent participant because of the anatomical developments occurring with the onset of puberty (Adirim & Cheng, 2003). Androgens begin to circulate in the male athlete and result in the development of greater mass and speed (and therefore power) (de Loes et al., 2000). Female athletes experience peak muscle strength when peak height velocity is reached; for boys, this peak muscle strength appears to occur after peak height velocity, which is approximately 6-12 months later than for the girls (Adirim & Cheng, 2003).

Knee injuries are 3 times more likely to occur in competition than in practice (Ingram et al., 2008). Researchers have attempted to estimate the risk of injury for athletes based on gender, and sport participation by collecting data on pre-existing injuries. Researchers defined athlete exposures as either a practice or competition in which an athlete could sustain an injury. Ingram et al. (2008) conducted a study to determine the number of injuries that occurred from the 2005-2007 academic school years. Certified athletic trainers reported 1383 knee injuries occurring over 3,551,131 athlete exposures (3.89 knee injuries per 10,000 athlete exposures). Of these injuries, 226 required surgery. The female athletes sustained 380 total knee injuries, of which 85 injuries required surgery for injury rates of 3.11 and 0.71 per 10,000 athlete exposures respectively. The male athletes sustained 1023 total knee injuries, of which 141 injuries required surgery for injury rates of 4.29 and 0.60 per 10,000 athlete exposures respectively. The most common diagnoses for major knee injuries requiring surgery were complete ligament tears (65.5%), torn cartilage (20.3%), incomplete ligament tears (6.0%), and fractures (2.6%) (Ingram et al., 2008).

Powell and Barber-Foss (1999) also looked at these same 9 sports, and reported injury rates twice those reported by Ingram and colleagues (2008). Decreases in injury rates could be from multiple factors over a 10-year period as a result of new injury prevention techniques, and improvement in diagnosis and treatments. In addition, injuries, which could have sidelined a player in the past, now can be treated with minimal or no time lost to the athlete. Although each study saw different injury rate percentages, the trends remain similar and have indicated female athletes are at an increased risk of knee injury. Injury rates in the Powell and Barber-Foss (1999) study also reported knee injuries were 3 times more likely to occur during competition than practice.

Female high school athletes were nearly 2 times more likely to require surgical intervention than were injured male athletes (Powell & Barber-Foss, 1999). In addition, Ingram et al. (2008) discovered girls' basketball players were more likely to require surgical repair for the knee injury sustained (36.3%). The female basketball athletes were the most likely to sustain season ending knee injuries (25.8%), followed closely by girls' soccer (23.7%), and volleyball (23.0%). For this study, the most common injury mechanism was described as contact with another person (44.2%), followed by no contact (36.6%), contact with the playing surface (15.5%), illness (1.7%) and contact with a playing apparatus (1.5%).

Knee surgeries account for upwards of 60% of the sports-related surgeries required every year (Powell & Barber-Foss, 1999). For many athletes, knee injuries pose a significant cost (Arendt & Dick, 1995) and require substantial rehabilitation visits (de

Loes et al., 2000; Louw, Manilall, & Grimmer, 2008). Traumatic injuries requiring reconstruction of the ACL are also commonly associated with increased risk of early onset osteoarthritis (Ingram et al., 2008).

Financial Implications

American's participation in sports has increased dramatically over the past few decades. With the number of participants on the rise, researchers and clinicians have seen a corresponding increase in the number of reported injuries to the lower extremities, especially the knee. Research indicates similar numbers of knee ligament sprains occur in females and males prior to the onset of puberty; however, females have higher rates immediately after the growth spurt associated with maturation. It has been estimated that 80,000 or more ACL tears are sustained by Americans alone. Approximately 50,000 ACL reconstruction surgeries are performed each year; however, some individuals opt out of the invasive surgical procedure and are therefore limited in their sports participation either through restriction or activity or functional bracing (de Loes et al., 2000). Shea and colleagues (2004) analyzed the insurance claims of 6 million youth soccer athletes and discovered a peak in the number of claims reported by female athletes and their families beginning around age 12.

Injury to the ACL typically involves lengthy and costly rehabilitation protocols, and to a relatively high extent, is followed by different degrees of impairment (Fagenbaum & Darling, 2003; Koon & Bassett, 2004). The consequences accompanying the initial acute injury can result in traumatic effects in one's personal or professional life, and their role in society, particularly as a member of a successful athletic team at a club, high school or university (de Loes et al., 2000; Fagenbaum & Darling, 2003; Koon & Bassett, 2004; Sell et al., 2006). ACL injuries have been associated with the highest percentage of permanent disability for all sport injuries (Caraffa, Cerulli, Projetti, Aisa, & Rizzo, 1996; Ford et al., 2003; Koon & Bassett, 2004). Some studies have suggested many athletes sustaining an ACL injury have an increased risk of suffering from osteoarthritis at some point during their lifetime as well (Ford et al., 2003).

The study by Shea and colleagues (2004) suggested ACL injuries can occur in children as young as 5 years of age. The group of scholars indicated the youngest female filing a claim during the studies parameters was for a 12-year-old girl. In a five-year period (starting in 1995) 8215 insurances claims were submitted to Bene-marc, Inc. a company that provides insurance policies for adolescent soccer athletes throughout the United States (Shea et al., 2004). A disproportionate number of claims (37%) were attributed to ACL injuries in the female athletes, whereas the claims for males only represented 24% of total claims (Shea et al., 2004). The scholars also indicated the number of injury claims appeared to increase at age 11-12 for both females and males until age 18 (Shea et al., 2004).

The physical cost of injury is blatantly obvious and tangible, however, the financial cost of injury is something few families or individuals think about prior to sport participation and ultimate injury. The financial obligation to correct issues associated with knee ligament injury represent a significant percentage of expenditure for the treatment of high school and collegiate athletes by private insurances as well as institutions of higher learning and their collegiate athletic programs. Across the United States of America, patients are spending an estimated $17,000 in healthcare costs such as surgical intervention and rehabilitation resulting from ACL injury (Childs, 2002; Ford et al., 2003; Hewett et al., 1999; Griffin et al., 2000; Koon & Bassett, 2004). In total, the cost associated with ACL treatment is over a billion dollars, with the annual expenditure for high school and collegiate female athletes reaching over $646 million alone (Chappell et al., 2005; Childs, 2002; Grace, Sweetser, Nelson, Ydens, & Skipper, 1984).

These financial figures are compounded when assessing the cost of the initial care of all ACL injuries or the conservative management and/or rehabilitation of patients who choose not to undergo ACL reconstructive surgery. In nearly 50-70% of all cases, patients with ACL injuries also had meniscal injuries requiring surgical attention (Bach et al., 1997; Bach et al., 1983). In addition, these numbers fail to take into account the future economic impact of injury such as the cost of treating long-term complications of the posttraumatic degeneration which results in many patients who sustain injury to the ACL, even those who undergo ligament reconstruction (Griffin et al., 2000). In general, female athletes had higher total costs for knee injury treatment and rehabilitation than did their male counterparts in sports such as soccer, downhill skiing, ice hockey, alpine skiing, and basketball (de Loes et al., 2000). The substantial economic burden is disproportionately placed on the female athlete and possibly her family. This financial encumbrance is further aggravated via the traumatic effects from potential loss of seasonal sport participation, future athletic scholarship funding, professional earnings, as well as the generalized effects on the athlete's mental health and possible academic performance (Ford et al., 2005; Ford et al., 2003; Toth & Cordasco, 2001). At least 8 months of post-surgery rehabilitation is needed for most athletes to return to full unrestricted return to play sports activity (McAllister et al., 2003).

In addition to the economic and psychological issues commonly associated with traumatic life-altering event in the lives of young athletes; rupture of the ACL also caries the risk of physiological deficit, which must be addressed by medical personnel when attempting to rectify the unstable knee.

Risk Factors for Noncontact ACL Injury

Past research has studied several possible risk factors associated with noncontact ACL injuries (Anderson et al., 2001; Piasecki et al, 2003). Neuromuscular risk factors have been reported to be one of the most likely explanations for the discrepancy in injury rates between the genders (Anderson et al., 2001; Chappell et al., 2005). Female athletes perform athletic maneuvers in a manner that could predispose them to higher risk of ACL injury. Research suggests muscle fatigue decreases dynamic knee stability, which could expose the female athletes to an increased incidence of knee injury (McAllister et al., 2003; Rowe et al., 1999). An increase in tibial translation was documented following an isokinetic fatigue protocol of the quadriceps femoris and hamstrings muscles (Hahn et al., 1999; Rowe et al., 1999). ACL risk factors can be described in terms of intrinsic and extrinsic factors to better elucidate the phenomenon associated with gender and the risk of ACL injury.

Intrinsic factors (*Figure 2.2)* include internal or anatomical differences, which could contribute to the increased injury rates, based on gender (Baker, 1998; James et al., 2004). Intrinsic factors are usually described as fixed or resolute because they are relatively unchangeable and are generally genetically predetermined in a person at birth. Intrinsic factors include but are not limited to: (1) increased knee joint laxity, (2) hormonal influences, specifically sex hormones, (3) femoral notch size, (4) ligament size, and (5) lower limb alignment including quadriceps angle (Arendt et al., 1993; Arendt & Dick, 1995; Chappell et al., 2005; Gwinn et al., 2000; James et al., 2004; Kirkendall & Garrett, 2000; Swartz et al., 2005; Toth & Cordasco, 2001; Wojtys et al., 1996).

Extrinsic factors (*Figure 2.2)* include the surrounding environment, which could contribute or influence injury rates predisposing an athlete to the risk of injury (Baker, 1998; James et al., 2004). Extrinsic factors are described as more mutable, and can to some extent, be modified through hard work and time. Extrinsic factors include but are not limited to: (1) level of conditioning, (2) muscle strength, (3) altered motor control strategies, (4) skill level, (5) playing surface and environmental conditions, (6) coaching differences, (7) muscular imbalances and (8) experience (Arendt et al., 1993; Arendt & Dick, 1995; Chappell et al., 2005; Gwinn et al., 2000; James et al., 2004; Kirkendall & Garrett, 2000; Swartz et al., 2005; Toth & Cordasco, 2001; Wojtys et al., 1996). Extrinsic factors, especially the association between altered motor control strategies and increased rates of injury in female athletes, have not been studied with any intensity, despite the dynamic nature of athletic injuries (Arendt & Dick, 1995; Chappell et al., 2005; Swartz et al., 2005).

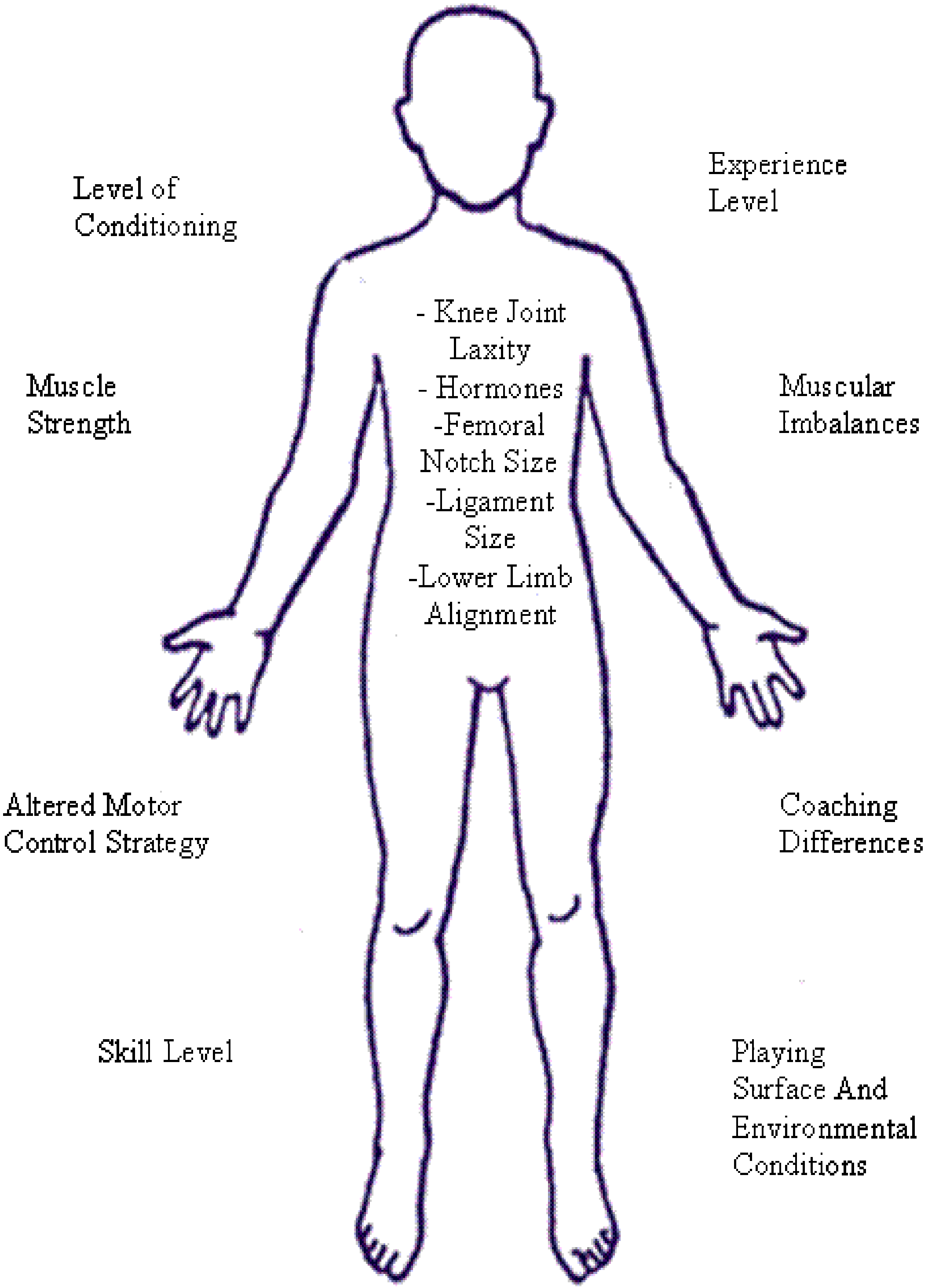


Figure 2.2. Internal And External Factors Affecting The Female Athlete

Previous research has separated inquiries into approximately four categories: environmental, hormonal, anatomical and neuromuscular. These four categories are addressed below.

Environmental Risk Factors

Researcher have studied prophylactic knee braces and shoe surface interfaces as potential causative environmental factors associated with the increased risk of knee injuries. Prophylactic knee braces may provide structural support to the knee during athletic competition (Bieze, 2004; Griffin et al., 2000). In addition, researchers have hypothesized that the interaction of shoes, including cleats, with an athletic playing surface could have the potential to cause injury to the athletic knee (Griffin et al., 2000).

Bracing

Functional and prophylactic knee braces were developed and introduced in the late 1970s to decrease the risk and reduce the severity of injury to the ligaments of the body (Griffin et al., 2000). Prophylactic braces protect and, in theory, prevent healthy athletic participants from injury to the ACL or the MCL (Prentice, 2009). Functional braces have also been designed to protect the surgically reconstructed ACL during athletic activity as well. Braces traditionally encapsulate the thigh and calf with fabric, metal, and/or plastic fasteners. Knee braces are frequently custom-molded with constraints to theoretically control rotational stress and/or tibial translation and effectively limit knee extension during a stop, jump and land (Bieze, 2004; Prentice, 2009). Many braces of recent design are lightweight and are hinged to allow for greater ranges of motion, or can be restricted to prevent ranges of motion based on protocol (Prentice, 2009).

Functional knee braces can be used to help provide support to an unstable or strength-decreased knee following injury during return to play activity (Bieze, 2004). Functional braces are designed to restrict terminal knee extension and improve the stability of surgically repaired ACL extremities (Prentice, 2009).

Many reports have indicated that prophylactic braces successfully decreased the number of injuries sustained by high school and collegiate athletes; however, as time passed, the statistics did not remain consistent. Upon further examination, many studies actually indicated the number of knee injuries increased while athletes were wearing prophylactic braces (Griffin et al., 2000). The protective effectiveness of knee braces is extremely controversial (Greene, Hamson, Bay & Bryce, 2000; Starkey & Johnson, 2006). In the mid 1980s, a position statement was created and released by the American Academy of Orthopaedic Surgeons stating that the evidence of the proposed injury reduction benefits for athletes using prophylactic and functional bracing is inconclusive at best and suggests medical personnel administer braces with discretion instead of mandates requiring individual or team use (Griffin et al., 2000). For many practitioners, braces are generally accepted to have little to no effect on functional performance measures, the reduction of knee ligament injuries, or the improvement of functional performance and are therefore not used; however, the practice of prophylactic bracing athletes has yet to be completely terminated at all levels of competition (Prentice, 2009).

The effectiveness of prophylactic braces still remains a mystery. Many coaches still insist their athletes wear such braces during competition; however, the braces might pose as much problem as they do effectiveness. It is also important to examine other environmental factors such shoe-surface interface when attempting to identify risk factors for lower extremity injury.

Shoe-Surface Interface

Athletic maneuvers such as running, quick stops and starts, and rapid change of directions result in excessive horizontal forces between the shoe and the environmental playing surface (McClay, Robinson, Andiacchi, et al., 1994b). Research has suggested cutting and shuffling movements could magnify the magnitude of transverse (sideways) forces equaling or exceeding the athlete's normal bodyweight (McClay, Robinson, Andiacchi, et al., 1994a; McClay et al., 1994b). An athlete must have a high traction coefficient between the shoe and surface to prevent slipping (Shorten et al., 2003). The amount of traction a player can create can help determine the extent to which a player can "lean" into a cutting movement without slipping (Page, 1978). When the shoe contacts a playing surface, a certain amount of friction created to allow for athletic movement. Increased levels of friction from the interaction allow the person to quickly accelerate, decelerate, or change directions and are generally associated with better athletic performance. As the level of coefficients of friction increases, however, so does the risk of injury (Griffin et al., 2000).

Many sports require athletes wear cleated shoes such as football, soccer, baseball, softball, and track while they interact with different types of surfaces including grass, dirt, artificial turf, synthetic fields, or clay based track surfaces. Torg and Quedenfeld (1971) were some of the first researchers to delineate the role a shoe plays with the playing surface. This research has contributed to understanding and preventing noncontact lower extremity injuries. Most of the research concerning shoe surface interfaces has examined the effect of cleats on different turf potentially reducing injury mechanisms in sports such as football, soccer, lacrosse, rugby, and others. The number and size of cleats on a shoe have been correlated with the number of injuries sustained to the knee and ankle in American football; Torg & Quedenfeld (1971) concluded that cleats with fewer spikes paralleled a reduction in injury rates. To determine the relative incidence of ACL tears, researchers implemented four different cleat designs utilizing 3119 high school football athletes between 1989 and 1991 (Lambson, Barnhill & Higgins, 1996). The researchers concluded knee injuries were most likely to occur in athletes wearing cleats with long, irregular spikes placed primarily on the peripheral margin of the sole, with a number of smaller, pointed cleats positioned interiorly on the sole (Lambson et al., 1996). In addition, the researchers indicated athletes were more likely to sustain injury on natural grass than on other surfaced for the subjects in this study. Football athletes have recently been using soccer-style shoes during their sport participation (Prentice, 2009).

Football cleats have been designed and tested in laboratories with the cleat construction, arrangement and architecture factoring into the torsional resistance encountered by the foot and lower leg (Griffin et al., 2000). When an athlete attempts to change direction or perform athletic maneuvers, the lower extremity could be placed at risk for injury if the cleated foot is planted and gets stuck (i.e., either encounters a large amount of friction and is stopped in its movement or becomes embedded in the ground as the athlete attempts to move). Shoe designs have changed over the years to optimize performance. A shoe-surface design might be created with less frictional forces and thus presumably a “safer” environment; however, it may not allow for optimal sporting performance (Griffin et al., 2000). In fact, if the athlete cannot produce enough friction to complete athletic maneuvers, the risk of injury increases as well. Decreased frictional forces occur within the environmental factors of athletic participation, for example if an athlete attempts a change of direction on wet surfaces or slick surfaces, then the potential for slipping and incurring injury is elevated.

Ultimately, the choice of an appropriate shoe use on any given playing surface requires a personal judgment regarding the level of risk (either of slip or of foot fixation) determined acceptable. In an additional study, Torg and colleagues (1974) attempted to quantify the injury potential of friction created in shoe surface interface combinations be calculating a “release coefficient” as determined by the peak torque developed at contact between the shoe and given playing surface (Torg, Quendenfeld, & Landau, 1974). The amount of traction between a sport shoe sole and an athletic playing surface is essential for performance and the safety of an athlete. Traction is deemed necessary to athletic performance; however, excessive traction increases the risk of “foot fixation,” an etiological factor in some sports injuries (Shorten, Hudson, & Himmelsbach, 2003).

Many factors contribute to the probability of slipping or foot fixation including: shoe and surface properties, athlete dynamics and the variations introduced by competitive contact sports (Shorten et al., 2003). The coefficients of traction on the turf surface can also be affected by temperature, moisture or contaminants, and age of the surface (Bowers & Martin, 1974; Shorten et al., 2003; Torg, Stilwell, & Rogers, 1996).

It is difficult to ascertain the specific and limited influence of environmental factors in conjunction with ACL injuries. Past research exploring the benefits and detriments of prophylactic and functional knee bracing and shoe surface interfacing have encouraged the identification of possible risks of injury and have opened the door to some injury prevention methods; however, no such study has demonstrated a direct correlation to the predisposition of knee ligament injury with specific environmental factors. As more shoes are manufactured and shoes are used for different sports (even some different than the manufacturer's intended use), the risk of injury becomes more uncertain. More research is needed to determine the effects of shoe-surface interface and athletic related injuries for sports requiring cleats, such as football and soccer.

The interaction between the shoe and the surface is of much concern to athletes and health care professionals because of the reciprocal relationship between traction and safety. Further research is necessary to determine the effect of specific types of shoes on surfaces and their relation to lower extremity injury.

Hormonal Risk Factors

The investigation into the role of sex hormones and risk of injury is an area of active inquiry as a potential causative explanation for the increased risk of injury to the female knee. Evidence suggests the hormonal fluctuations occurring during the menstrual cycle could influence knee joint laxity and muscle stiffness in female athletes (Shultz, Kirk, Johnson, Sander, & Perrin, 2004). Puberty signals the flood of secondary sex hormones into the body and is introduced with an associated increase in the secretion of hypothalamic gonadotropin-releasing hormones, which stimulates the secretions of the gonadotropin luteinizing hormone (LH), and follicle stimulating hormone (Liu et al., 1996). The follicle stimulating hormones act on the ovaries and testes to stimulate the secretions of sex hormones (Aagaard, Simonsen, Magnusson, Larsson, & Dyhre-Poulsen, 1998). In addition, researchers have identified estrogen and progesterone receptor sites in the human ACL cell increasing the evidence of the impact of hormonal fluctuations and the risk of knee injury (Griffin et al., 2000; Liu et al., 1996). The establishment of sex hormone receptors within the knee ligament was a significant find and could indicate female sex hormones could play a vital role in the explanation of the gender biased injury rates (Griffin et al., 2000).

Females experience normal reproductive hormonal changes throughout the body with an average menstrual cycle lasting approximately 28 days (Harmon & Ireland, 2000). The levels of hormones in the bloodstream fluctuate throughout the menstrual cycle, and the menstrual cycle has been divided into three subsections based on the state of the ovary: follicular, ovulatory, and luteal phases (Harmon & Ireland, 2000). The follicular phase is initiated on the first day of the menstrual cycle and usually lasts for approximately 9 days. In the follicular phase, estradiol and progesterone are at their lowest levels (Boden et al., 2000b). Around days 10-13 of the menstrual cycle, luteinizing hormones rise substantially to initiate ovulation, resulting in estradiol reaching peak levels and progesterone drops to its lowest level. A rush of luteinizing hormones is released approximately 24 hours prior to ovulation (Boden et al., 2000a; Boden et al., 2000b). The ovulatory phase is sustained for approximately 5 days for the average female. The luteal phase encompasses approximately day 15 of the cycle to the end of the cycle (estimated at 14 days). The luteal phase has been associated with a slow decline in the availability of luteinizing hormone, increases in progesterone, and it appears that estradiol begins to rise again during this time (Boden et al., 2000a; Boden et al., 2000b; Harmon & Ireland, 2000). Researchers have postulated that estrogen has the ability to relax soft tissue which could predispose the female knee to increased risk of injury during specific times in the menstrual cycle (Boden et al, 2000a; Boden, 2000b). The highest levels of estrogen have been found to occur in the days just before ovulation during the ovulatory phase, suggesting women could be at an increased risk during the ovulatory phase (Boden et al, 2000a).

The menstrual cycle has been studied to determine if certain phases have been associated with increased risk of injury to the female athlete. The ovarian sex hormones have been associated with possible tissue alterations and increased incidences of ACL injuries (Slauterbeck et al., 2002; Wojtys et al., 2002a; Wojtys et al., 2002b). Wojtys and colleagues (2002a, 2002b) conducted a study using sixty-nine women who incurred an acute noncontact ACL injury. The athletes were asked to give two urine analyses, one within 24 hours of the injury and a second sample within 24 hours of the first day of her next ovulatory cycle. The researchers reported more injuries occurred in the ovulatory phase of the menstrual cycle (approximately days 10-14) and fewer injuries occurred in the follicular phase (days 1-9) of the cycle (Wojtys et al., 2002a; Wojtys et al., 2002b).

In direct contrast to Wojtys and colleagues study (2002a, 2002b), Arendt and colleagues (1993) found the ovulatory phase to the least likely time of injury (Arendt et al., 1993). Research is inconclusive and incomplete at this time in determining the specific menstrual phase most likely to predispose a female athlete to injury. Slauterbeck et al. (2002) performed a study to determine if menstrual histories provided by athletes at the time of injury could be confirmed via measurements of salivary estrogen levels to determine if ACL injuries were occurring more frequently during specific phases of the menstrual cycle. The study established a correlation between self-reported last menstrual period at the time of injury and the actual salivary estrogen and progesterone levels at r = 0.95 (a significantly high correlation) for the 21 athletes who provided information to this research. Again in contrast to the Wojtys and colleagues' study (2002a, 2002b), this study determined females were more likely to tear their ACL during the follicular phase of the menstrual cycle rather than the ovulatory phase (Arendt et al., 1993; Slauterbeck et al., 2002). Recently researchers have suggested the likelihood of sustaining an ACL injury is more prevalent during the preovulatory phase of the menstrual cycle than the postovulatory phase (Arendt, Bershadsky & Agel, 2002; Beynnon et al., 2006; Myklebust et al., 2003; Slauterbeck et al., 2002; Wojtys et al., 2002a; Wojtys et al., 2002b).

Estrogen and relaxin, female sex hormones, fluctuate during the menstrual cycle in association with other secondary sex hormones in the female body (Toth & Cordasco, 2001). These hormones are suggested to contribute to decreased ligament strength or result in altered muscle recruitment resulting from cyclic changes in female hormones possibly contributing to the increased rate of injury in female athletes (Ford et al., 2003; Hewett, 2000; Hewett et al., 1999). Estrogen receptors have been identified in the ACL and in skeletal muscle (Lemoine et al., 2003; Wilk et al., 2005), and could contribute to the strength and stiffness created by cellular metabolism (Huijing and Jaspers, 2005). Estrogen has been demonstrated to alter the composition and mechanical properties of the

ACL in the female knee. Estradiol may decrease the tensile properties of tissue and decrease the rate of collagen synthesis (Toth & Cordasco, 2001). Relaxin is another sex hormone associated with injury in the female knee. Relaxin is typically found in pregnant women, which allows for pelvic changes to accommodate fetal passage through the birth canal, and may increase the risk of ACL injury in pregnant women (Moeller & Lamb, 1997).

In addition, the authors suggested that women who use oral contraceptives appeared to be at a diminished risk for ACL injuries during the menstrual cycle. Oral contraceptives help regulate the female's hormone production to maintain a consistent cycle, and reduce the amount of hormonal shift, which occurs in non-medicated women (Wojtys et al., 2002b). Literature has suggested that a relationship could exist between peaks in estrogen levels and increases in the laxity of the ACL (Deie, Sakamaki, Sumen, Urabe, & Ikuta, 2002; Shultz et al., 2004; Slauterbeck and Hardy, 2001). Changes in the tolerance capabilities of the ligament could predispose the ACL to injury at tensile loads lower than normal or could alter the protective muscular reflex actions occurring in conjunction with ACL tissue receptor stimulation (Raunest, Sager, & Burgener, 1996). Tension loading of the ACL is regulated by the muscular system, which attempts to limit the external forces and moments created through bony motions (Dedrick et al., 2008). Estrogen receptors have been discovered in skeletal muscle and thus could provide an influence on neuromuscular control and myofascial force transmission pathways (Huijing and Jaspers, 2005; Lemoine et al., 2003; Wilk et al., 2005).

The relative time for the muscle around the hip to activate (hip muscle onset timing) may occur with different frequencies based on the phases of the menstrual cycle

(Dedrick et al., 2008). According to the research, a 33 ms difference was noted for the onset timing of the hip musculature, specifically the gluteus maximus and semitendinosis, between the early follicular and luteal phases. Cowling and Steele (2001) have suggested delayed semimembranosus muscle onset in women during ground reaction could implicate a muscle synergy pattern including delayed quadriceps contraction when compared to hamstring contract, placing the knee at risk for injury. The causal relationship between hormonal fluctuations and potential for ACL injury remains an elusive mystery for researchers. As of yet, no direct evidence has been presented to determine if ACL injury is more likely to occur in a specific phase of the menstrual cycle.

At this point in time, it is difficult to say exactly what role the female secondary sex hormones play in the process of ACL injury; however, many scholars have studied the risk of injury. Along with the hormonal differences, suggested anatomical differences occur between men and women possibly leading to injury in the lower extremity.

Anatomical Risk Factors

Anatomical and structural differences among the genders have been an area of popular research because the variables are tangible and appear to be an obvious visual source of differences for injury. Anatomical differences between males and females could play a significant factor in the incidence and type of knee injuries sustained during athletic participation (Griffin et al., 2000). Female knees tend to have similar physical attributes such as: increased joint laxity, which is the combination of joint hypermobility and musculotendinous flexibility; increased quadriceps angle (Q-angle), which the angle created by the shaft of the tibia and shaft of femur; smaller ACL ligament size; and smaller femoral intercondylar notch size than their male counterparts (Beighton, Solomon, & Soskolne, 1973; Beynnon et al., 2005; Ford et al., 2003; Harmon & Ireland, 2000; Hertel, Dorfman, & Braham, 2004; Jansson et al., 2004; Moeller & Lamb, 1997; Nguyen & Shultz, 2007; Bultman, Wellink, & van Dongen, 1997; Robinson, Sanchez-Ballester, Bull, Thomas, & Amis, 2004; Rosene & Fogarty, 1999; Scerpella, Stayer, & Makhuli, 2005; Shultz et al., 2005a; Shultz et al., 2005b; Toth & Cordasco, 2001; Uhorchak et al., 2003). No differences have been observed in adult male and female's measurements of tibial torsion (Nguyen & Shultz, 2007), navicular drop, (Hertel, Dorfman, & Braham, 2004; Nguyen & Shultz, 2007; Trimble, Bishop, Buckley, Fields & Rozea, 2002) or rearfoot angle (Astrom & Arvidson, 1995; Nguyen & Shultz, 2007).

Compared to males, greater anterior pelvic tilt has been documented in adult females (Hertel et al., 2004), in addition with other factors such as hip anteversion (Nguyen & Shultz, 2007); tibiofemoral angle (Nguyen & Shultz, 2007); and genu recurvatum (Nguyen & Shultz, 2007; Trimble et al., 2002). In addition, other contributions that could expose the source of ACL injuries include the experience level of the participant in competition and the size of the athlete who is competing, as women tend to be smaller in body size than are males (Moeller & Lamb, 1997; Toth & Cordasco, 2001).

Researchers have studied the material landscape of the ACL and determined ACL volume; cross-sectional area and material properties of the ACL are highly correlated with gender, height, age and weight (Anderson et al., 2001; Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006; Chandrashekar, Slauterbeck, & Hashemi, 2005). The ability of the ligaments to sustain pulling forces (stretch) and rebound to preexisting levels without damage is referred to as laxity (Shultz, Nguyen & Schmitz, 2008). General joint laxity has been identified as a causative factor in the increased risk of overall injury in sports. When the ligaments are lax, they allow for greater movement and thus could increase the ultimate ability for shearing forces to occur to the knee during dynamic movement. The ACL in the female athlete is less elastic and fails at a lower level load (lower failure strength), even after researchers adjust for age, body anthropometrics, and ACL size (Chandrashekar et al., 2006). Several studies have indicated that women's hormonal levels vary in conjunction with a cyclical menstrual cycle variations and are associated with changes in genu recurvatum (the hyperextension angle of the knee) and general joint laxity in the healthy female knee (Shultz et al., 2008).

Researchers are uncertain at this time whether clinical laxity differences have a significant impact on weight bearing neuromechanical activity, however, most researchers would concur with the statement that women generally have more joint laxity than men (Toth & Cordasco, 2001). Researchers have theorized that women do not have the ability to control their joints due an increase in the laxity of the ligaments ultimately resulting in a less stable joint (Godlfuss et al., 1973). Males on the other hand, have less laxity and therefore have more stable joints, and could have the ability to better control their kinematic joint angles. In female athletes, a mild increase in both anterior and posterior knee laxity (18-20% increase) has been reported when running activities have been performed for more than 30 minutes (Griffin et al., 2000; Godlfuss et al., 1973). The increased laxity effects appear to be transient in nature and laxity returns to normal limits approximately 60 minutes after activity cessation (Griffin et al., 2000; Godlfuss et al., 1973). Most sports activity requires the athlete participate in running activities for longer than 30 minutes and it therefore stands to reason most female knees are subjected to increases in the laxity of the ligaments during exercise, possibly allowing for enough movement in the knee to result in trauma or injury.

Another area of interest concerns the quadriceps angle (Q-angle) as a potential contribution to the increased risk of knee injury for the female athlete. The Q-angle is formed by the intersection of two lines: the first line follows the anterior superior iliac spine on the anterior surface of the hip to the center of the patella (flowing down the shaft of the femur), and the second line runs from the center of the patella to the tibial tubercle (flowing down the shaft of the tibia) (Prentice, 2009). Angles are typically measured with the athlete positioned in a long-sit, however, researcher have described measuring the Q-angle in a standing or short-sitting position as well. Angles measured up to 17° in the long-sit position are considered normal in females, and angles up to 10° are considered normal in the male population (Moeller & Lamb, 1997). Females tend to have higher Q-angles due to the reproductive nature of the pelvic structure of the female body (Starkey & Johnson, 2006). The female's hips are generally wider than male hips, and therefore, the line of pull for the femur is generally larger for females than males. Data collected during research has confirmed the relationship between the Q-angle and patellofemoral tracking issues and injury, however, studies have yet to reveal a direct relationship between Q-angle and ACL injury (Moeller & Lamb, 1997).

The female's femoral notch angle is generally smaller than their male counterparts, and the femoral notch height is larger, which could impact the femoral notch impingement theory (Chandrashekar et al., 2005). Even when body size, body composition, and height are taken into account, it appears women tend to have smaller intercodylar notch widths than do males (Harmon & Ireland, 2000). Femoral notch width is proven to be a good predictor of ACL size (area and volume) in males but the same theory does not correspond to females (Chandrashekar et al., 2005). Femoral intercondylar notch width and ACL size have been studied extensively as anatomical variables and gender-specific risk factors. The ACL has been suggested to be at a greater risk of injury when housed in a small intercondylar notch, commonly described as an A-shaped notch. With a smaller I-notch, it is possible that the ACL is more likely to come in contact with the medial femoral condyle when the knee is in flexion, and to impinge on the anterior notch when the knee is in full extension (Toth & Cordasco, 2001).

When examined radiographyically, individuals who present with a decreased intercondylar notch width are at an increased risk of knee ligament injury (Anderson et al., 2001; Chandrashekar et al., 2006; Chandrashekar et al., 2005) When reviewing the literature, there appears to be some discrepancy between researchers report gender differences and those who report little to no anatomical differences between the genders (Moeller & Lamb, 1997). Lombardo and company (2005) suggested the risk of sustaining a noncontact ACL injury in professional basketball players was not correlated with the size of the intercondylar notch (Lombardo, Sethi, & Starkey, 2005). As a result of the questionable conclusions in research surrounding the notch width argument, many studies have since focused on the size of the ligament within the notch. The results of Lombardo and company's study (2005) could suggest anatomical predispositions and/or changes, which have occurred naturally or have been developed by the professional athlete and as such researchers have been cautioned against the application of these findings to other populations.

Harmon & Ireland (2000) reported a smaller notch width might house a smaller, thinner, and weaker ACL, which would be less able to withstand the same amount of force as larger ligaments. New evidence infers females with smaller femoral notches also present with smaller ACL's and therefore could have inferior structural integrity when compared to similar male athletes (Shultz et al., 2008).

A prospective study was used to evaluate the associated risk of injury among United States of America military cadets (Uhorchak et al., 2003). The study found men with smaller intercondylar notches and associated generalized joint laxity had a much higher probability for injury (nearly 7.8-fold increase of injury). For the female cadets, factors such as narrow notch size, increased body mass index, and associated general joint laxity placed them at risk for injury. Another important finding suggested women who had knee laxity values more than one standard deviation away from the mean were nearly 2.7 times more likely to sustain injury (Uhorchak et al., 2003).

The female ACL is smaller in cross-sectional area, volume and length compared to the male ACL, even when researchers account for body anthropometry (Chandrashekar et al., 2005). When adjusting for age and body anthropometrics, the ultrastructural analysis of the ACL shows the collagen fiber area percentage (area of collagen fibers/total area of the micrograph) is smaller in females than in males (Hashemi et al., 2008). According to Chaudhari and colleagues (2008), the average ACL volume for ACL injured athletes was nearly 8% smaller than the average ACL volume for the control group (1955 mm^3 and 2117 mm^3, respectively). The study indicated 16 of the 27 injured participants had a smaller ACL size than their matched control. Similar results have been found when looking at the differences in ACL volume between men and women

(Chandrashekar et al., 2005). ACL volume could play an essential role in identifying risk of injury for athletes (Chaudhari et al., 2008).

Anderson and colleagues (2001) studied the correlation of anthropometric measurements and postulated gender difference risk factors and determined male athletes were generally heavier than female athletes with less overall body fat than the women in the study. MRI measurements were taken to determine the condylar width, notch width and ACL area. Males appeared to have significantly greater condylar widths, notch widths, and ACL areas. When the researchers adjusted the measurements for weight, the mean ACL area for men was significantly greater than women. In addition, Anderson and colleagues (2001) also addressed the correlation between ACL area and height of the athlete, which demonstrated for males as the height increased, the size of the ACL increased; however, the same was not said for the female athletes. This finding indicates taller women in this study did not show any change in ligament size than the shorter women, possibly indicating taller women might have an increased strain placed on a weaker, thinner ACL ligament whereas, as males ACL size increased with height, and therefore, the their ligament's tensile strength increases in accordance (Anderson et al., 2001).

Other researchers have indicated the probability of injury decreases during contact sports as the bodyweight of the participants increases (Hutchinson & Ireland, 1995). The likelihood for women to sustain a serious injury is in a competition with male counterparts increases; however, when participants of the same gender and size compete, the likelihood of injury is somewhat diminished (Hutchinson & Ireland, 1995). In contact sports, participants of mismatched size or decreased skill level place the novice player, or smaller sized player, at an increased risk of injury (Hutchinson & Ireland, 1995). This research theory has been demonstrated as a causative factor for the increased risk of female collegiate military members at the Naval Academy and the American Military Academy. Separate studies completed by Gwinn and colleagues (2000) and Uhorchak and colleagues (2003) determined women were more likely to sustain a serious injury while participating in intramural activities at military academies. The intramural activities provide an equal playing field in which both genders compete together on coed teams (Gwinn et al., 2000; Uhorchak et al., 2003). In addition, the women tended to play less important positions either via self-selection or through team placements (Gwinn et al., 2000). As well, the researchers postulated the women might not have engaged in as aggressive play as they might have where they playing against other women (Uhorchak et al., 2003).

Anatomical and structural differences among the genders could identify several obvious differences within the structure and base of support for athletes. The bodies for male and female athletes are built to function in very specific ways; however, to date it is unclear if the anatomical differences are the main source for risks of injury. It appears the female knee has some very specific attributes, and those attributes could contribute to an athlete's posture and running mechanics.

Posture

Studies have identified posture as an important factor, which could be contributing to injury. A more vertical posture, or more erect posture, during gait, landing and cutting activities can be created by greater extension angles in the trunk, hip and/or knee position, and could contribute to the risk of ACL injuries by increasing the vertical landing forces sustained to the subject during locomotion (Griffin et al., 2000). Differences between the genders suggest females are more likely to display a more erect posture during athletic activities than are their male counterparts which could help account for the disparaging reports of female ACL injury rates (Decker et al., 2003; Huston et al., 2001; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Salci, Kentel, Heycan, Akin, & Korkusuz, 2004).

The stress and strain of the lower extremity is directly influenced by the alignment of the ligaments, muscultendinous structures and knee compartments (Decker et al., 2003). Alignment differences have been correlated with an increased development of patellofemoral disorders, but as of yet, have failed to be directly linked to the increased risk of ACL injuries in the female athlete (Hutchinson & Ireland, 1995). Future investigation is necessary to determine the factorial influence of the anatomical structures of the lower extremity, gender and the risk of serious knee injury. Posture has been studied with minimal success to attempt to determine the risk of injury. Posture typically gets thrown into categories within the anatomical risk factor subgroup. Another subgroup, which has received much attention, is the neuromuscular risk factors associated with injury.

Neuromuscular Risk Factors

Neuromuscular activation patterns and gender have been a hot topic of debate among scholars and have been widely implicated as a potential causative factor in the increased risk of injury to the female knee. Investigators have suggested different factors as possible contributors suggested to increase risk of knee joint injury; one such contributor is a neuromuscular activation pattern. Muscles contract in response to a stimuli or series of stimuli to create neuromuscular differences and subsequential activation patterns. Implicated differences between the genders have been studied to determine the length of time needed to produce specified force muscle levels, with the females consistently performing tasks significantly slower than their male counterparts (Huston & Wojtys, 1996; Rozzi et al., 1999).

In recent reviews of literature, researchers have identified neuromuscular control as one of the most influential and most modifiable factors in the risk of injury to the ACL of the female knee (Griffin et al., 2000). “Neuromuscular control” refers to the sensory stimuli responding to unconscious activation of the dynamic restraints surrounding a joint (Griffin et al., 2000). Neuromuscular control can also refer to any aspect regarding the nervous system regulation of muscle activation and the factors contributing to athletic task performance (Riemann & Lephart, 2002).

Neuromuscular control is essential to avoid extreme or hazardous joint positions resulting in injury. Previous studies have indicated female athletes perform athletic activities, such as cutting, with knee joint mechanics possibly predisposing the athlete to risk of injury (Malinzak et al., 2001; Pollard et al., 2004; Sigward & Powers, 2006). Females in these studies have performed tasks less flexed (Malinzak et al., 2001; McLean et al., 2004b), more abducted (Malinzak et al., 2001; McLean et al., 2004a; McLean et al., 2004b), and with increases in internal knee adduction (McLean et al., 2004a) during stance.

A study by Hewett and company (1999) provides integral insight into the effects of neuromuscular training on the female knee injury rates for female athletes participating in high-risk sports. Non-trained female athletes appeared to more susceptible to knee injuries and performed athletic maneuvers differently than the athletically trained male and female groups, and even the non-trained male group, which could be a factor predisposing women to injury (Hewett et al., 1999). This study indicates the incidence of serious knee injury was 2.4 to 3.6 times higher in the untrained group than the trained group, depending on whether the sport of volleyball was included along with the soccer and basketball analysis. Since this study, several other researchers have also suggested women, especially athletically trained women, appear to perform athletic maneuvers in a manner which exposes the knee joint to large amounts of ligament strains (Chappell et al., 2002; Colby et al., 2000; Hewett et al., 2005; Malinzak et al., 2001).

Increased vertical landing forces, muscle strength, and firing patterns have been suggested as potential predispositions for serious knee injury (Colby et al., 2000). Rozzi and colleagues (1999) considered the neuromuscular characteristics of male and female basketball and soccer players to determine differences in movement characteristics (Rozzi et al., 1999). Thirty-four healthy collegiate level athletes who were on the varsity basketball or soccer teams were used as participants in this study. The study utilized a testing device designed to measure the detection threshold for passive motion. The unit moved the knee joint into flexion and extension through the axis of the joint, while a rotational transducer provided angular displacement values among other balance, EMG, laxity and proprioceptive assessments. The research suggests women performed tasks with inherently greater knee joint laxity values, produced significantly greater electromyographic (EMG) peak amplitude of the lateral hamstring when landing from a jump, and denoted significant periods of time to detect the knee joint laxity values when compared to similar male counterparts (Rozzi et al., 1999). Female athletes appear to adopt a compensatory muscle-activation pattern to achieve functional joint stability when performing athletic maneuvers (Riemann & Lephart, 2002; Rozzi et al., 1999). Neuromuscular control impacts factors including muscular imbalances, muscle recruitment patterns, and movement execution patterns.

The neuromuscular recruitment patterns of the thigh muscles surrounding the knee are responsible for providing stiffness and dynamic stability at the knee during locomotion (Solomonow, Baratta, & D' Ambrosia, 1989). The muscles surrounding the knee joint (quadriceps and hamstring muscles) undergo a preparatory and reflexive co-contraction to prevent injury by increasing joint stiffness (Baratta et al., 1988). Researchers such as Huston & Wojtys (1996) and Rozzi et al. (1999) have indicating female subjects utilize alternative muscular activation and recruitment strategies and require substantial time to produce an equivalent muscle force as male analogue subjects (Huston & Wojtys, 1996; Rozzi et al., 1999).

Neuromuscular recruitment patterns were analyzed by Hewett (2000) to determine if differences exist between or among the genders. The researchers discovered female athletes tend to contract their quadriceps more rapidly and with an increased force in direct response to an anteriorly directed force to the back of the calf. In opposition, the male athletes primarily respond to this provocative anterior tibial translation by first contracting the hamstrings. The hamstrings function as a posterior muscle group creating an agonist (resisting force) restraint to protect the knee from too much anterior shear force, thus reducing the strain placed on the ACL during anterior tibial translation (Hewett, 2000). The quadriceps muscles act as an antagonist, at knee flexion angles of less than 45°, which significantly increases the strain placed on the ACL (Hewett, 2000). Results from this study and other similar studies suggest female athletes tend to be primarily “ligament-dominant” in their joint positioning strategies, whereas males tend to rely on “muscle-dominant” strategies to stabilize the joint during locomotion (Bolgla, 2008; Hewett, 2000; Houck & Yack, 2003). A majority of the literature has focused on the shear forces acting on the knee as a result of neuromuscular recruitment patterns (De Carlo, Irrgang, Wilk & Rothstein, 2000; Houck & Yack, 2001; Hurwitz, Andriacchi, Bush-Joseph, & Bach, 1997). From these findings, scholars have implemented neuromuscular intervention strategies by attempting to modify “high-risk” movement patterns (Hewett et al., 1996; Myer, Ford, & Hewett, 2005). These studies have suggested retraining and modification of lower limb movement control is possible through training paradigms (Hewett et al., 1999; Myer et al., 2005).

Huston and Wojtys (1996) compared the neuromuscular activation strategies for elite male and female athletes and non-athletes. An isokinetic dynamometer was implemented to test the strength of the athletes and non-athletes to demonstrate the slower generation of peak knee flexion (hamstring) torques in the female athletes compared to their male counterparts. The researchers postulated the female athletes would have significantly increased quadriceps strength, and diminished hamstring strength, thus resulting in an anterior translation of the tibia. The group of female athletes generated maximum knee extension torque prior to maximum knee flexion torque. None of the non-trained female, non-trained male and athletic male groups demonstrated this phenomenon during research testing. Both female groups (non-athletes and athletes) exhibited statistically diminished quadriceps and hamstring muscle strength at the 60°/second speed compared with both male non-trained and athlete groups, even when the researchers accounted for (normalized) body weight. The research indicates female athletes could activate different muscular recruitment strategies than non-trained females, and both of the male groups, which indicates training and conditioning during sports activities could result in predilection of knee injuries. It appears females place their knee in a valgus force via employment of muscular contraction patterns during semi-static joint positions (Ford et al., 2005). This valgus position increases the knee loads experienced by the subject and could magnify the knee joint torques during dynamic maneuvers as a result of inappropriate muscular contractions (Ford et al., 2005).

Several of the musculature surrounding the knee also crosses the hip joint, and as such scholars have suggested neuromuscular imbalances could occur in the hip as well. Poor neuromuscular control of the hip may contribute to the increased risk of injury at the knee. Bolgla (2008) examined the hip and knee neuromuscular activity between males and females and found females not only demonstrated a quadriceps dominant pattern, but also activated the vastus medialis sooner than males. The study also observed an increase in femoral internal rotation and adduction (both of which have been shown to increase strain on the ACL). The gluteus maximus and gluteus medius muscles attempt to control these muscles. The scholars found no statistically significant gender differences in amplitude; however, males tended to demonstrate a significant delay in vastus medialis activation relative to the gluteal muscles studied. Earlier and more efficient hip muscle activation could decrease the forces imparted throughout the lower extremity during impact. Consequently, the amalgamation of the quadriceps dominance and lack of hip control could further increase the risk of ACL injury for the female athlete.

Trunk Flexion

Researchers have hypothesized increases in trunk flexion during landing activities could result in corresponding greater knee and hip flexion, decreases in knee valgus, hip internal rotation and hip adduction angles (Blackburn & Padua, 2008). ACL loading could be increased during landing as a result of the concomitant increases in hip and knee flexion. These altered lower extremity kinematic findings could identify a potential risk for knee injury. Blackburn & Padua (2008) suggested ACL injury prevention programs should focus on training athletes to land from jumping maneuvers with an increased trunk flexion to alter the kinematics of the lower extremity.

Proprioception

Balance can be operationally defined respectively in static and dynamic contexts as the ability to maintain a base of support with minimal movement or as the ability to perform tasks while maintaining a stable position (Winter, 1990). Challenging the sensoriomotor system could enhance balance and provide a way to train the athlete’s body to recognize itself in space and provide a way to decrease injury when the body responds to incorrect body positioning. Balance is influenced by somatosensory, visual, and vestibular information obtained simultaneously from the body and from the motor responses affecting coordination, joint range of motion (ROM), and strength (Palmieri-

Smith, McLean, Ashton-Miller, & Wojtys, 2008; Palmieri, Ingersoll, Stone & Krause, 2002).

Researchers have noted differences in ankle and knee proprioceptive abilities between trained athletes and a control group (Aydin, Yildiz, Yildiz, Atesalp, & Kalyon, 2002; Lephart, Giraldo, Borsa, & Fu, 1996). Some researchers have indicated increases in balance and proprioceptive abilities are the result of repetitive athletic experiences influencing the motor responses (Balter, Stokroos, Akkermans, & Kingma, 2004). Others argue enhanced balancing ability is the culmination of training experiences, which influence an athlete's ability to distinguish and attend to relevant environmental and visual cues (Ashton-Miller, Wojtys, Huston & Fry-Welch, 2001).

The sensoriomotor systems influencing the balancing abilities of the trained athlete depend on the demand of the skill requirements and environmental challenge (Bressel, Yonker, Kras, & Heath, 2007). Gymnasts perform tumbling, jumping and landing maneuvers with bare feet on surfaces varying in surface composition. A gymnast is required to perform skills with exaggerated joint ROM, strength and coordination (Bressel et al., 2007). Basketball players typically perform upper extremity dominant passing, shooting and dribbling skills while running, jumping, and landing on flat, stiff surfaces while wearing athletic shoes (court shoes, crosstrainers, or running shoes). The skill demand of basketball players requires high amounts of joint accelerations from jump landings and cutting maneuvers (McClay et al., 1994a; McClay et al., 1994b). Football and soccer athletes are required to perform tasks while wearing cleated or noncleated shoes on variable authentic, artificial and synthetic turf conditions (Orchard, 2002). The skill sets required for soccer are dominated by the lower extremity requiring the athlete to perform passing, shooting, and dribbling skills. The environmental demands, athletic skill sets, and proprioceptive awareness and ability vary based on the demands of the sport (McClay et al., 1994a).

In 2007, a team of researchers examined the proprioceptive abilities of female athletes who participated in different sports and determined female basketball athletes demonstrated inferior static balancing abilities when compared to gymnasts and soccer athletes (Bressel et al., 2007). The researchers used the BESS (Balance Error Scoring System) to quantify the amount of static proprioceptive abilities available to the subjects in the study and suggested gymnasts might have had an advantage in this test because they perform daily tasks requiring static balancing skills during beam and floor routines that basketball and soccer athletes typically do not participate in. The researchers suggest, basketball players are rarely required to balance motionless on a single leg, and are more often required to focus attention on outside stimuli such as the movement of the ball and player position cues (Bressel et al., 2007).

Bressel and colleagues (2007) also examined dynamic balance by the Star Excursion Balance Test (SEBT), which requires athletes to stand on one leg and reach out in one of eight directions around a circle to touch marks on the ground with the toe of the other foot. The researchers determined the female soccer athletes had the best scores, which could be due to the requirements of the sport demanding a player perform single-leg reaching movements outside of the base of support. Soccer athletes are frequently required to stand on one leg and kick or receive a pass with another leg; therefore, dynamic one-leg stabilization is a requirement of the sport. Results such as these suggest balance training and proprioceptive training are essential in developing a well-rounded athlete. More research is needed to determine the effects and correlations between proprioceptive abilities and the risk of injury.

Muscle Activation

The way athletes activate muscles could determine the landing characteristics associated with issues contributing to the risk factors for injury (Rozzi et al., 1999). When athletes cannot control their joints through muscular activation, the body is forced into positions often resulting in trauma and injury. Experts suggest active muscle stiffness could contribute to leg stiffness thus increasing landing forces as a result of performing tasks with more extended extremities (Farley, Blickhan, Saito, & Taylor, 1991; Farley & Gonzalez, 1996; McMahon & Cheng, 1990).

Research is divided when trying to determine the muscle activation differences between the genders. A study (Rozzi et al., 1999) indicated that compared to males, female basketball and soccer athletes increased hamstring activity (a peak in EMG amplitude following initial ground contact) when performing a single-leg landing. In opposition, Fagenbaum & Darling (2003) suggested no lower extremity muscle EMG activation differences were apparent between males and females during landing. Malinzak et al. (2001) was one of the first researchers to suggest female athletes activate the quadriceps at elevated levels during side-step cutting tasks. Cowling and Steele (2001) indicated a delay in the hamstring muscle activation onset compared with females during landings with a single leg. These reports suggest females could activate inappropriate muscle activity strategies relative to the forces acting on the knee during landing. Rozzi et al. (1999) also reported female collegiate athletes had increased hamstring amplitude during single-leg landings.

Bolgla (2008) supported the “female quadriceps dominant” activation pattern stating females activate the vastus medialis sooner than males. Activation of the quadriceps during dynamic activity could result in excessive anterior tibial shear strain and therefore place stress on the ligaments of the knee. Increases in the adduction and internal rotation of the femur can also increase the strain of the ACL and could be controlled by the gluteal muscles (Bolgla, 2008). Bolgla (2008) found no significant differences in gluteus maximus and medius amplitudes; however, males demonstrated a significant delay in vastus medialis muscle activation in relation to the gluteus maximus and medius. Athletes who activate hip muscles earlier in the movement appear to have better control in the stabilization of the hip and dampen the valgus moment at the knee (Cowling & Steele, 2001). In addition, increased hip stability might decrease the force attenuation sustained to the lower extremity during high impact movements. Any decrease in strength surrounding the hip, thigh, or knee could expose the knee to an increased risk of injury, particularly in the female athlete (Bolgla, 2008).

The ability of muscle attenuation during landing and cutting has been an area of recent research (Chappell et al., 2002; Colby et al., 2000; Cowling and Steele, 2001; Lephart et al., 2002a; Lephart et al., 2002b; Malinzak et al., 2001). Females may land with the lower extremity in an extended position (Decker et al., 2003), which requires the hamstrings to fire in response to anterior translation of the tibia (Lephart et al., 2002a; Lephart et al., 2002b). In recreationally active athletes, researchers have indicated females demonstrate greater quadriceps activation and decreased hamstring activation compared with males when performing hopping, cutting and lunging maneuvers (Hanson, Padua, Blackburn, Prentice, & Hirth, 2008) whereas Malinzak and colleagues (2001) utilized a recreational athletic population. In a study in 2008 by Hanson and colleagues, gender differences were identified between Division I collegiate soccer athletes with females exhibiting increased vastus lateralis and gluteus medius activation amplitude during side-step cutting (Hanson et al., 2008). The authors from this study hypothesized these findings support the quadriceps dominant theory for female athletes performing athletic tasks such as side cuts, crosscuts, and straight runs (Hanson et al., 2008).

Large external loads are applied to the knee joint during landing requiring the muscles surrounding the joint to help function as anatomical moment arms to reduce the potential for ligamentous loading (Andriacchi, Andersson, Ortengren & Mikosz, 1984; Lloyd & Buchanan, 2001). The central nervous system is better able to facilitate a detailed muscle activation protocol and adjust muscle activation patterns when destabilizing forces were anticipated by the athlete (Branch, Hunter, & Donath, 1989). The musculature providing dynamic support to the knee is required to support the ligaments by providing resistance to the anterior translation of the tibia on the femur occurring after the landing and helping to protect the ACL (Cowling & Steele, 2001). The researchers suggested differences in hamstring activation timing between genders, which could be responsible for variations in the ACL injury rates among the genders (Cowling & Steele, 2001). Researchers have indicated postural adjustments might be preprogrammed and could provide insufficient time for the female athlete's central nervous system to plan appropriate muscle activation strategies to counter and stabilize the joint while loads increase during unanticipated cutting tasks (Besier et al., 2001).

Side cutting and crosscutting maneuvers have been utilized in female athletes to demonstrate the women had less hamstring muscle activity (lateral hamstring and medial hamstring) and increased rectus femoris activity (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007a; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007b). Landry and colleagues (2007a; 2007b) performed one of a handful of studies utilizing an elite adolescent soccer population. The authors indicated sagittal plane kinematic (flexion/extension) differences exist between male and female athletes at the hip. Female athletes demonstrated far less hip flexion angles than male subjects during side-cut maneuvers and no differences in knee kinematics (Landry et al., 2007a; Landry et al., 2007b). Another study reported similar results for hip kinematics (McLean et al., 2004a; McLean et al., 2004b) and no differences in knee kinematics between the genders (Ford et al., 2005; Pollard et al., 2004; Sigward & Powers, 2006). Other studies do not report similar findings, as Pollard et al. (2004) suggested differences exist in hip flexion angles between the genders but not at the knee. Malinzak et al. (2001) and McLean et al. (2004a, 2004b) indicate female athletes have less knee flexion during side-cut maneuvers than do male athletes.

The difference in the rate of injury for genders appears to explode during maturation. It has been suggested children employ movement strategies that do not expose them to injury, and somewhere along the developmental line, this strategy is altered, changed, modified, or lost (Russell, Croce, Swartz, & Decoster, 2007). Negative relationships have been discovered between the number of years an athlete has participated in activity and the co-contraction ratio of the quadriceps and hamstrings muscles (Russell et al., 2007). In the Russell (2007) study, the co-contraction of the anteroposterior (quadriceps and hamstring muscles) has been discovered to occur differently between the young and older subjects when preparing for athletic maneuvers. The adults demonstrated more muscle activity from the hamstring group in relation to the quadriceps group (in particular, the vastus medialis). It appears children demonstrate a strategy employing the larger and stronger muscles of the hip and torso to control the forces sustained during landing as opposed to the adults who appear to activate a knee or ankle joint strategy to control the deceleration forces of landing (Russell, 2007).

Researchers have postulated highly trained athletes could implement different neuromuscular strategies in the quadriceps, hamstrings, and hips to decelerate from a single-leg landing (Viitasalo, Salo, & Lahtinen, 1998). In the mid 1970s, researchers tested the hip and thigh strength of 119 subjects presenting with lower extremity injuries such as ankle and knee ligament injuries, patellar pain, and knee arthritis and discovered specific weakness in the muscles surrounding the injury, specifically in the hip abductor and adductor (Nicholas, Strizak, & Veras, 1976). These muscular differences have remained relatively under-investigated until the mid-1990s. The body and its movement within space is of particular concern when creating and implanting training regiments to prevent injury and should continue to be an area of focus for current and future research.

Muscular Imbalances

Research has suggested athletes who have muscular imbalances between their quadriceps and hamstring muscles are at an increased risk for injury (Chappell et al., 2002). Muscular imbalances can lead to dysfunction and incorrect motion of the body. When the muscles on either side of a joint are too strong, the body ends up developing a compensatory movement to account for the strength imbalances, which requires other structures of the body to perform tasks beyond or in addition to normal (implicit) functions.

<u>Quadriceps to Hamstrings Ratio.</u> Muscular imbalances between the quadriceps and hamstrings muscles may contribute to the risk of injury for athletes (Chappell et al., 2002). Research has been devoted to determine the gender impact and the differences in thigh muscle (quadriceps and hamstrings) strength and some investigators have postulated females become reliant on ligamentous stability of the knee whereas males tend to rely more on muscles (particularly the hamstrings) to provide joint stability during functional motor movements (Hutchinson & Ireland, 1995; Solomonow et al., 1987; Toth & Cordasco, 2001).

Previous research has indicated noncontact ACL injuries are more likely to occur at or near the point of foot strike during athletic maneuvers. At this particular point in an athlete’s movement execution, the quadriceps muscles are eccentrically contracting to resist the resultant knee flexion at foot strike which results in the maximum amount of muscular force (Delfico & Garrett, 1998). The noncontact ACL injury mechanism has also been characterized with deceleration; change of direction as in cutting or landing; and a varus/valgus moment about the knee; or an internal/external rotation of the leg during physical motion (Colby et al., 2000; Wojtys et al., 1996).

The role the quadriceps muscles play in ACL injury mechanism has been of particular concern for researchers because the quadriceps muscles have been implicated in their role in pulling the tibia anteriorly which results in stresses placed on the ACL at low knee flexion angles (Colby et al., 2000; Griffin et al., 2000; Hunter et al., 2003; Kirkendall & Garrett, 2000). The hamstrings act in antagonistic opposition to the pull of the quadriceps and provide support for the ACL; the hamstrings pulls the tibia posteriorly on the femur and provides dynamic stability to the knee by resisting both mediolateral and anterior translational forces on the tibia (Colby et al., 2000; Hewett, 2000; Hewett et al., 1999; Hunter et al., 2003; Lephart et al., 2002a; Lephart et al., 2002b).

Overactivation of the quadriceps muscle could create an anterior shear force of the femur on the tibia possibly causing the athlete to rely on ligamentous support rather than muscular restraint to slow the body's trajectory of movement (Ahmad et al., 2006; Lephart et al., 2002a; Lephart et al., 2002b; Rosene et al., 2001). Recent research has implicated the hamstrings muscles to act as a counterbalance to attempt to decrease the directed shear force placed on the tibia in relation to the femur (Ahmad et al., 2006; Rosene et al., 2001; Solomonow et al., 1987). In 2000, Aagaard and colleagues studied the antagonist hamstrings moments during locomotion. The study suggested the hamstrings had the ability to counteract the anterior tibial shear force and excessive internal tibial rotation induced by the quadriceps when the knee nears terminal extension (Aagaard et al., 2000). Co-activation of the hamstrings has been suggested to assist the mechanical and neurosensory function of the anterior cruciate ligament in the knee during athletic movement (Aagaard et al., 2000; Osternig et al., 1995). The ACL is aided in maintaining the knee joint stability via coactivation of the antagonist hamstrings musculature during active knee extension by exerting an opposing torque to the anterior tibial displacement induced by the quadriceps muscles (Osternig et al., 1995). Researchers have suggested female athletes tend to perform activity with a dominant quadriceps strength and muscle-firing pattern compared to their male counterparts. Studies have indicated athletes of different sports have exhibited specific quadriceps:hamstrings ratios, indicating adaptations in muscle activity might be altered as a result of varied levels of competition and training (Hewett, 2000; Rosene et al., 2001; Wojtys et al., 1996).

Huston and Wojtys (1996) were among the first researchers to indicate females are more quadriceps dominant than are their male counterparts (Baker, 1998; Ford et al., 2003; Hewett et al., 1996; Huston & Wojtys, 1996; Toth & Cordasco, 2001). This finding suggests females tend be reliant on the quadriceps muscle more than the hamstrings muscle to functionally stabilize the knee during locomotion (Huston & Wojtys, 1996). Males tend to activate the hamstrings muscles at three times the level of females during landing, jumping and cutting maneuvers. Female athletes also appear to have significantly less lower-extremity endurance and strength, particularly when examining the hamstrings (Aagaard, Simonsen, Andersen, et al., 2000; Colby et al., 2000; Hewett, 2000; Lephart et al., 2002a; Lephart et al., 2002b; Toth & Cordasco, 2001). Some researchers have suggested the possible use of quadriceps and hamstrings strength ratios as a screening tool to identify potential predisposition to injury during high school and collegiate pre-participation physical examinations (Rosene, Fogarty & Mahaffey, 2001).

In 2006, researchers deliberated on the effects of gender and maturity on the quadriceps:hamstrings ratio and found female participants who were considered “mature” (beyond adolescence) had significantly greater ratios when compared with the immature girls, immature boys and mature boys (Ahmad et al., 2006). Ideally, a person would have a relatively low quadriceps:hamstrings ratio suggesting the quadriceps muscles are just as strong as the hamstrings muscles; however, these findings suggest the mature women's quadriceps muscles are significantly stronger than their hamstrings muscles, a finding that was unique to this group only. Researchers have suggested females increase their quadriceps strength much more after the onset of menarche than their opposing hamstrings muscle strength. This phenomenon could possibly increase the risk factor for anterior cruciate ligament injury (Ahmad et al, 2006).

Men demonstrate faster generation of peak hamstring muscle torque compared to women (Wojtys & Huston, 1994). Huston and Wojtys (1996) suggested female athletes have weaker quadriceps muscles, even when researchers adjust for bodyweight differences between the genders. Other scholars have suggested the ratio for strength between the quadriceps and hamstrings groups are much lower for males than females, suggesting the quadriceps muscles in females are much stronger than the hamstrings muscles (Anderson et al., 2001).

A female who is quadriceps dominant tends to contract the quadriceps muscle in response to anterior tibial translation in contrast to the male athletes who tended to contract their hamstrings muscles in response to tibial translation (Ahmad et al., 2006; Hewett, 2000; Griffin et al., 2000; Rosene et al., 2001; Wojtys et al., 1996). When the hamstrings are neglected or forgotten during training or the quadriceps can become over-trained, and results in increases in quadriceps muscle imbalance could potentially have a detrimental effect on a player's performance, possibly leading to injury (Aagaard et al., 2000; Osternig, et al., 1995).

Colby et al. (2000) performed a study using fifteen healthy seasoned (collegiate) athletes and recreational athletes (nine men and six women) to determine the muscular activity of the quadriceps and hamstrings muscles during cutting maneuvers. When the authors analyzed the four athletic cutting maneuvers (sidestep cutting, cross-cutting, stopping, and landing), the researchers determined the contraction of the quadriceps muscles while the knee was flexed from 0° to 30° resulted in contraction levels creating significant anterior shear forces on the proximal tibia. These findings could indicate the force of the eccentrically contracted quadriceps was greater than the tensile strength of the ACL, thus predisposing the female athlete to risk of injury (Colby et al., 2000). The eccentric contraction produced by the quadriceps could be capable of creating an anterior shear force capable of rupturing the ACL in a noncontact athletic injury mechanism (Colby et al., 2000).

A "normal" hamstrings:quadriceps ratio is 50% to 80% averaged through a full range of motion, with lower ratios present at faster speeds (Buchanan & Vardaxis, 2003; Dunnam, Hunter, Williams, & Dremsa, 1988; Hewett et al., 1999; Rosene et al., 2001). This ratio suggests the hamstrings have approximately 50%-80% of the capacity of the quadriceps muscle strength with higher percentages representing faster Isokinetic speeds (Rosene et al., 2001). Researchers have hypothesized quadriceps:hamstrings ratios exceeding 60% could place the athlete at an increased risk of ACL injury (Aagaard et al., 2000; Hewett et al., 1999). This means athletes who exhibit 60% or greater strength in the quadriceps over the hamstrings muscles, could be placing themselves at an increased risk of injury. Female athletes, and to an even greater extent, female non-athletes, are at an increased risk of quadriceps:hamstrings muscular imbalances resulting in ACL injury (Malinzak et al., 2001). As an athlete's muscular imbalance is rectified and the ratio approaches 100% (Ratio of 1:1; quadriceps:hamstrings ratios become closer to equal strengths), the hamstrings increase their functional capacity to stabilize the knee in dynamic situations (Ahmad et al., 2006). Increased knee stability could decrease the possibility of anterolateral subluxations of the tibia on the femur (Rosene et al., 2001). Debate has surfaced among researchers when data have attempted to quantify quadriceps:hamstrings ratios as a predispositional factor in knee injuries in females (Buchanan & Vardaxis, 2003).

It appears the quadriceps-to-hamstrings ratio can be affected when examining medial and lateral forces for females. Scholars tend to think this medial-to-lateral difference could cause a limited ability to resist loads from abduction during locomotion (Palmieri-Smith et al., 2008). The ACL can be strained with associated higher abduction loads possibly leading to injury. Muscular imbalances between the medial and lateral aspect of the thigh should be corrected to attempt to reduce the incidences of ACL injury in the female athlete (Palmieri-Smith et al., 2008). The quadriceps and hamstrings muscles are not the only muscles in the body that can have an imbalance. Any body system requiring an agonist and antagonist muscle to activate surrounding a joint to maintain stability can sustain an imbalance in strength such as the gluetal muscles and the adductor group.

Gluteals. The gluteal muscles, in particular the gluteus medius, are essential in providing muscular control of the hip motion during athletic tasks and as such, researchers have suggested these muscles could play an integral role in explaining the gender bias in injury prevalence (Hart et al., 2007). The gluteus medius, according to Anderson & Pandy (2003), is the primary abductor of the hip, assists in the control of femoral internal rotation during activity and provides pelvic support during the midstance of gait (Hart et al., 2007). In addition, the gluteus medius controls the multi-planar motion of the hip joint kinematics (Ireland, 1999; Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007). Improper activation or reduction of muscle activity in the gluteus medius could result in less resistance to internal rotation and adduction of the hip; both positions have been linked with an increased risk of ACL injury (Ireland, 1999). In the open kinetic chain, intermediate and posterior gluteus medius fibers are activated to induce hip abduction and lateral rotation, respectively (Delp, Hess, Hungerford & Jones, 1999). In theory, the eccentric contraction of the gluteus medius intermediate and posterior fibers occurs during the deceleration phase of a closed kinetic chain activity and would function to control the hip adduction and internal rotation range of motion (Carcia & Martin, 2007).

Russell et al. (2006) performed a study analyzing the single-leg drop landing and found no differences in gluteus medius activation between the genders (Russell, Palmieri, Zinder, & Ingersoll, 2006). Zazulak et al. (2005) examined the gluteus medius and gluteus maximus activity in 13 female and nine male NCAA collegiate athletes (Division I) during a single leg drop landing protocol and determined females had significantly lower gluteus maximus contraction levels, but no differences were noted in gluteus medius activation. In opposition, Hart and colleagues (2007) discovered the average gluteus medius muscle activity was significantly higher in males than females while performing a forward jump activity.

Carcia & Martin (2007) examined the surface electromyography (sEMG) activity before and after a drop jump protocol between genders. The authors suggested looking at pre-landing activity as a measure of anticipatory muscle contraction. Post-landing muscle activity allows the researcher to examine how a muscle responds to demands from forces and torques as a result of the landing forces (Carcia & Martin, 2007). The gluteus medius activity was not statistically different for the pre-landing and post-landing muscle activity or for the genders. Although the authors did not find statistical differences between the genders, the female subjects exhibited greater amounts of variability in gluteus medius EMG activation levels.

Many researchers do not test healthy hips with isokinetic dynamometers because of the high level of torque that the hip can produce, many times exceeding the dynamometer's torque limit (Shultz et al., 2005a). Due to these constrictions, many times, the hip is not isokinetically tested. It has been suggested, "a more functional examination of hip strength can be gained using free weights and weight machines" (Shultz et al., 2005a, p. 494). Researchers need to continue to probe further to determine if a specific quadriceps:hamstrings strength ratios to be used as a screening tool to prevent ACL injuries during medical pre-participation physical examination screening processes.

The hip has presented difficulty in assessment and the diagnosis of issues because of the overall complexity and strength requirements at the hip during dynamic movement. It appears differences could exist at the hip, which could expose the knee to increased risk of injury. Differences in the way athletes train for given sports could create these muscle imbalances, and therefore, should be cautioned against overtraining in a single plane of movement.

Sports Differences

Researchers have suggested that the choice of sport participation to which an athlete commits could be a primary risk factor in injury ratios. Authors have determined differences exist between trained and matched controls when examining ankle and knee proprioception through challenges to the sensoriomotor system enhancing balance (Aydin et al., 2002; Lephart et al., 1996). New research seems to indicate differences in balance, proprioceptive abilities, and neuromuscular recruitment activation based on the sport in which an athlete participates (Cowley et al., 2006). Female high school basketball athletes injure their ACL more often while landing or jumping (60% of injury mechanisms), whereas females participating in soccer only documented 25% of injury mechanisms resulting from jumping or landing (Cowley et al., 2006). According to Cowley and colleagues (2006), the most frequent ACL injury mechanism for female soccer athletes appears to be related to the cutting or change of direction maneuver rather than jumping or landing.

Other researchers have suggested this phenomenon as well citing the primary physical requirements of each sport is dependent on the injury mechanism (Cowley et al., 2006; Moeller & Lamb, 1997; Powell & Barber-Foss, 2000). During basketball, athletes are required to jump and land while shooting, rebounding, and defending an opponent, whereas the primary requirements for soccer do not rely on jumping as much as the cutting and change of direction because soccer is dominated by the lower extremity and is played at a much more ground level (the ball stays on the ground and is kicked from person to person). The variation in injury mechanism potential could be due to the nature of the specific sport demands or to the developed or inherent neuromuscular strategy cultivated by the athlete (Cowley et al., 2006; Powell & Barber-Foss, 2000). Athletes are trained in specific movement patterns as predetermined by the requirements and demands of the participant's sport, thus indicating certain movement could place an athlete at risk for injury during normalized athletic pursuits. Researchers have suggested recently the major factor in the increased incidence of ACL injuries could be more dependent on sport-specific criterion rather than based on differences between the genders (Cowley et al., 2006; Powell & Barber-Foss, 2000).

Bressel and researchers (2007) examined the factors affecting balance in female gymnasts, basketball and soccer athletes. Each sport requires individual sensoriomotor processes to perform the given tasks necessary for sport participation. Gymnasts are required to perform dynamic movements such as leaping, tumbling and balancing while barefoot on surfaces of varying stiffness and texture. Basketball athletes focus their efforts on upper and lower extremity activities including running, cutting, passing, shooting, dribbling, and screening other players on stiff surfaces while wearing court, cross-training, or running shoes. Soccer athletes perform tasks mostly with the lower extremities including passing, dribbling, and shooting while wearing turf shoes or cleats to play on grass or artificial turf.

For the Bressel et al. (2007) study, female basketball athletes demonstrated subpar static balance abilities when compared with female gymnasts and soccer athletes. The authors suggest the sport of basketball rarely requires an athlete to balance motionless on a single leg while attending to the ball and other stimuli on the court. Basketball players are typically instructed to keep a base of support in an athletic stance while playing. Soccer athletes are typically asked to perform single leg stances as they kick, pass, or receive the ball to and from other athletes on the field and therefore have more experience with the given tasks.

The sports in which an athlete participates can stress different factors and result in decreased muscle force production and ultimate fatigue. Sports such as basketball, football, lacrosse, soccer and rugby are described as intermittent activities requiring the athlete to perform rapid and successive sprinting and resting, or submaximal activities. Sports such as these require athletes to perform long periods of low intensity exercise interspersed with shorter periods of high intensity exercise (Davis & Brewer, 1993). All of these sports require athletes to engage in rapid bursts of sprinting in addition to multiple changes of direction throughout a practice, game or match. Frequently jumping, landing, and cutting are common components of locomotion in the athletic endeavor as well. Recently researchers have suggested the proportion of these activities could have a significant effect on injury mechanism based on sport protocol, for example, basketball players are required to perform jumping and landing in greater successions, whereas soccer players tend to perform more cutting and change of direction activities thus implicating injury mechanism as related to sports (Cowley et al., 2006).

Basketball

With the implementation of Title IX over the past few decades, nearly ten times as many high school girls are participating in competitive sports compared to the numbers in 1972 (Agel, Evans, Dick, et al., 2007a; Agel et al., 2007b; Baker, 1998; Huston & Wojtys, 1996). The National Federation of State High School Associations data indicates between 1988 and 1998 the number of participants in boy’s high school sports increased nearly 10%, while participation rates for girls' high school sports exploded and indicate a nearly 40% increase (Heidt et al., 2000). Adolescent female athletes are participating in organized athletics and achieving improved levels of fitness. Despite these changes in fitness, experience, and participation researchers have yet to see a corresponding reduction in the rate of injuries. Over the past 10 years, increased levels of fitness have not translated into a reduced risk of injury for female basketball players (American Academy of Pediatrics, Committee on Sports Medicine and Fitness, 2000; Mihata et al., 2006).

Basketball is not classified as a collision sport in most leagues, but the game is typically played in an accelerated and aggressive manner with a correspondingly high incidence of injury (Meeuwisse, Sellmer, & Hagel, 2003). Basketball athletes are required to participate in repetitive jumping, running, and cutting activities in a practice or game. Athletes are also required to engage in long distance running and interspersed rapid sprinting bursts during practice or game situations.

In 2006, researchers compared the frequency of knee ligament injuries in occurring in athletes within the Women's National Basketball Association (WNBA) and the National Basketball Association (NBA) and found that while the overall frequency of ACL injury was low, the injury rate was still 1.6 times higher for the participants in the WNBA (females) than the NBA (males) (Deitch, Starkey, Walters, & Moseley, 2006). The authors suggested the gender discrepancy might be lower for the professional sports than for other age groups and leagues as a result of the attrition rates and previous premature termination of athlete's careers prior to entering the professional ranks (Deitch et al., 2006). Basketball and soccer are two of the most common team sports for youth participants within the United States. Therefore, it is important to examine the differences between the two sports to determine the exact requirements for individual athletes and determine the exact mechanism of injury.

Soccer

Soccer, or futbol (football) as it is officially referred to by the Fédération International de Football Association (FIFA) and most of the rest of the world, is generally considered to be one of the most popular games in the world with more than 240 million registered players in approximately 150 countries annually (Giza, Mithofer, Farrell, Zarins, & Gill, 2005; Junge & Dvorak, 2004; Sandelin, Santavirta, & Kiviluoto, 1985). Much of the research pertaining to soccer athletes has utilized adult or professional soccer athletes, with most studies using male subjects rather than female subjects. Junge and Dvorak (2004) studied adult male professional soccer players throughout an entire macrocycle (an entire year of training, including the pre-season, in-season, and post-season) to determine exposure-related injury incidences. The risk of injury appears to coincide with the increase in age and competition levels. When examining the injury rates for youth soccer players, the 17- to18-year old age group appeared to have similar or even elevated injury rates than their adult counterparts. The results from this data indicate the increase in level of physical play and aggressive tendencies performed with the experience level of the players could increase the risk of overall injury; and in particular, ACL injury risk.

Giza and colleagues (2005) published a study to document the incidence of injury occurring in women's professional soccer and determined over half of the participants were injured at some point during the first two seasons of league play. The researchers indicated 202 players were listed on the rosters of the eight teams playing in the Women's United Soccer Association (WUSA), and found that approximately 110 of the players were injured at some point during the first two seasons. A total of 173 injuries occurred in the 110 injured players with an overall injury rate incident of 1.93 injuries per 1000 player hours in the two-year period. When the researchers examined the incidence of injury during practice and games, and found the risk of injury was 1.17 and 12.63 per 1000 player hours, respectively. Of the injuries, only 16% of the injuries were classified as chronic, nearly 82% of the injuries were acute. Of the injuries sustained, over half of the injuries sustained occurred to the lower extremity. ACL injuries accounted for 4.6% of all injuries documented. The rate of incidence of ACL tears was 0.09 per 1000 player hours (practice 0.04, game 0.90) (Giza et al., 2005).

Soccer is considered to be a high demand, multiple sprint sport which requires the athlete to consistently exert short, rapid, bursts of energy, usually lasting from 5-10 seconds, of maximal or near maximal efforts during participation (Dawson, Fitzsimmons, & Ward, 1993). The soccer athlete must have the ability to endure long periods of low to moderate intensity efforts in addition to these high demand tasks. Most soccer players are required to run prolonged distances as the ball moves up and down the field in conjunction with the changes in ball possession. Research has indicated a soccer player could perform 100 or more sprints during the course of a typical soccer match (Davis & Brewer, 1993). Soccer also requires a person to perform successive accelerations, decelerations, jumps, lands, changes in direction, and pivoting throughout a game in response to the changing moments and movement of the opposition during a practice or game situation. The demands of functional participation in soccer have long been epitomized the potentially harmful movements and increased risk of ACL injuries (Davis & Brewer, 1993; Dawson et al., 1993).

Movement Execution Patterns

Researchers have indicated gender differences could exist when performing demanding athletic maneuvers such as cutting and landing from a jump, or changing direction. According to Wikstrom et al. (2004) women tend to land with the knee and hip in more extended positions and thus subject themselves to higher ground reaction forces per body weight during the impact of landing. Other researchers have suggested women land from freefalling jumps with increased hip adduction, hip internal rotation, and knee abduction angles (Pollard et al., 2004). In addition, females have also demonstrated greater knee valgus angles at ground contact than similar male subjects, suggesting the load placed on ACL increases in conjunction with knee valgus angles (Lephart et al., 2002a; Lephart et al., 2002b).

Researchers have conducted retrospective studies and corroborating analysis of videotaped injuries to determine the mechanism of injury for the noncontact ACL injury. Researchers have indicated the knee flexion angle at the time of injury is somewhere typically between 30° of knee flexion and full knee extension (Delfico & Garrett, 1998). The videotaped ACL injuries also affirm an abrupt but significant deceleration prior to a change of direction as a more potential injury action than a pivoting motion around a planted foot (Delfico & Garrett, 1998).

McNair and Prapavessis (1999) collected vertical ground reaction forces from two hundred and thirty-four adolescent subjects (154 males, 80 females) aged 13 to 19 years old. Each subject was required to jump from a box (0.30 meters in height) onto a force plate. The researchers indicate differences in peak vertical ground reaction forces were determined by gender, sports level and activities necessary to compete in sports participation. Data indicate males had higher overall ground reaction forces than did the female subjects. The researchers defined competitive athletes as athletes who participated in a competitive sport 4-7 times a week. The competitive athletes were found to endure higher ground reaction forces than did recreational athletes, who were defined as athletes who participate in an activity 1 to 3 times per week. In addition, the researchers denoted athletes who participated in activities requiring significant amounts of jumping and landing and statistically compared them to subjects who participated in sports that do not require these activities. The participants who engaged in excessive jumping and landing activities withstood higher peak ground reaction forces than did athletes who were not required to participate in jumping and landing activates for participation (McNair & Prapavessis, 1999).

Dufek and Bates (1991) studied the ground reaction forces placed on the body when landing from a height of 0.40 meters. The subjects were asked to drop down from a box and land on forceplates with minimal impact. The researchers predicate the mean ground reaction forces were 3.85 times the body weight of the three subjects studied. While this study did not have a substantial subject population, these results could suggest the lower extremities and the ACL in particular are subjected to very substantial amounts of force when landing from a freefalling jump.

The ground reaction forces between female athletes and non-athletes have been compared to determine the effects of training on the forces sustained during physical activity. Seegmiller and McCaw (2003) established the difference between competitive female gymnasts and recreational athletes, with the gymnasts sustaining significantly higher vertical ground reaction forces during drop landings than did the female recreational athletes. Repetitive substantial vertical ground reaction forces sustained by athletes could expose the knee ligaments to increased risk of injury. High-impact loads are modifiable factors changed through instructional and training modifications. Essentially, athletically inclined women need to be taught how to land with diminished forces or to transmit the forces throughout the body so the ligaments are not the primary factor slowing the deceleration of the body during landing activities (Seegmiller & McCaw, 2003).

Sell and colleagues (2006) submitted data to suggest certain jumping and landing tasks are performed with altered neuromuscular and biomechanical characteristics at the knee joint based on gender. The researchers recruited 35 healthy high school basketball players, 18 male and 17 female, to perform planned and reactive double legged stop-jump tasks in three different directions. The research indicated no significant gender differences were noted based on gender; however, the females performed both reactive jumps and jumps requiring a right leg dominant athlete to move to their left in an approach which could increase the stress and strain on the ACL and could perhaps place the knee ligament at risk for injury. Less knee flexion angles were noted for female participants when performing reactive jumps. The female athletes appear to alter their neuromuscular and biomechanical patterns for activities consisting of unplanned jumping and landing strategies. Unplanned movements are commonly required in athletic activities, with the primary focus of basketball scoring and cutting to get the ball by faking opponents into believing he or she is moving in one direction while the actual movement continues in another direction (Sell et al., 2006). When the athlete is playing defense, the player must respond quickly to movements made by an opponent to avoid leaving their mark open and allowing them to score for the other team. In addition, an athlete must perform rapid and unplanned movements to avoid other players in the vicinity as well as environmental obstacles while focusing attention on other events.

Athletic activity typically consists of prolonged activity resulting in fatigue. A study conducted by Wojtys & Huston (1994) suggested fatigue could be a primary factor in the increased anterior tibial translational forces placed on the athletic knee during locomotion. Fatigue of the lower extremity muscles during physical activity may retard the potential dynamic stabilization and resulting knee defense mechanisms. The muscle firing in each of the medial and lateral quadriceps muscles was diminished by approximately 40% after fatiguing exercises were performed. The firing rate for the hamstring muscle saw similar decreases (lateral hamstring, 30% reduction in firing; medial hamstring, 35% reduction in firing) during fatigue exercises. This research indicates the muscle firing capability is greatly attenuated with the introduction of fatigue and could suggest a possible depreciation of muscular stabilization resulting from fatigue could allow for an increased anterior tibial translation possibly placing the ACL at risk for injury (Wojtys & Huston, 1994).

Research currently suggests the most vulnerable moment consistent with injury occurs in conjunction with ground contact while landing, coupled with an awkward body position. A series of studies in the past six to eight years suggests intercollegiate female athletes have significantly different proprioceptive characteristics, muscle firing patterns, and landing strategies compared with their male counterparts (Lephart et al., 2002a; Lephart et al., 2002b). Research such as this suggests several underlying physiological mechanisms could be potentially responsible, or bear some responsibility for differences between and among the genders (Lephart et al., 2002a; Lephart et al., 2002b). Research as of yet has failed to primarily focus on the younger athletes, whether due to ease of subject recruitment or the difficulties associated with studying minor participants. Younger athletes are playing sports at extremely competitive levels and injuries to the ACL still consist of traumatic surgical and rehabilitation requirements. It is imperative for researchers to focus on the younger athletic population to determine if these physiological effects occur in the younger population consistently as it does for the collegiate athletes.

Athletic Maneuvers

ACL injuries are most often the result of a noncontact, unanticipated or perturbed mechanism, frequently occurring during the landing or stance phase of “high risk” sporting postures such as sidestepping (Andrews, McLeod, Ward, & Howerd, 1977; Besier et al., 2001; Colby et al., 2000; Griffin et al., 2000; McLean et al., 2004a; McLean et al., 2004b; McLean et al., 2003). Locomotion during sports participation is not always anticipated during game and practice situations. Rather movement typically requires the athlete to respond suddenly to external stimulus such as avoiding another participant or following the bounce or pass of a ball (McLean, Neal, Myers, & Walters, 1999;

Myklebust, Maehlum, Holm, & Bahr, 1998). Therefore, researchers suggest preplanned cutting maneuvers are not a true representation of the load applied to the knee joint during actual sporting situations (Besier et al., 2001).

Laboratory experiments typically require basic equipment including a force plate, 3D high-speed motion capturing systems to provide kinematic and kinetic descriptions, and occasionally electromyography (EMG) equipment. Researchers have acknowledged some of the discrepancies found with the attempts to mimic "athletic activity" within a laboratory setting (Landry et al., 2007a; Landry et al., 2007b). Besier and associates (2001) have recognized potential disparities in unplanned sidestep executions which resulted in increases in external varus/valgus and internal/external knee moments when compared with more discrete movements which didn't reflect game play. In addition, some researchers criticize research studies examining the potential links between hazardous sports movement and injuries by focusing solely on the biomechanics of the knee joint (McLean et al., 2004a; McLean et al., 2004b). ACL injury is a result of excessive loads being placed on the knee, which are unsupportable by the bony, and muscular structure, thus resulting in copious ligament loads ultimately resulting in failure (Landry et al., 2007a; Landry et al., 2007b). Excessive knee loading could be potentiated via abnormal neuromuscular control elsewhere in the lower extremity; therefore, studies only examining the knee could be missing potentially influential and complicating factors in the reporting of significant results (Besier et al., 2001).

In the hope of creating game-like situations by forcing athletes to execute decision making processes in a rapid, split second succession, most recent laboratory studies have devised light-guiding systems to increase the number of motor responses an athlete must choose from to perform a desired movement much like they would be required to do in a game situation (Besier, Lloyd, & Ackland, 2003; Besier et al., 2001; Ford et al., 2005; Houck, Duncan & De Haven, 2006; Pollard et al., 2004). Light guiding systems have been implemented in laboratory research to help recreate unanticipated maneuvers and researchers have suggested these protocols better replicate a true-game scenario (Landry et al., 2007a; Landry et al., 2007b). Although admittedly it is difficult to assess exactly how well these unanticipated laboratory maneuvers mimic game-like situations.

During athletic competition, the execution of athletic maneuvers, such as a sidestep tactic requires a spatial and a temporal randomness making “on-site” analyses of lower extremity motion and function virtually impossible (McLean et al., 2004a; McLean et al., 2004b). Unanticipated maneuvers are commonly re-created in laboratory settings to better control movement execution patterns and evaluation in an attempt to replicate a true unplanned game-like scenario (Besier et al., 2001; Ford et al., 2005).

Researchers have indicated unplanned activities require participants to increase the dynamic and functional stability of the joints exponentially when compared to pre-planned activities (Landry et al., 2007a; Landry et al., 2007b). Besier et al. (2003) demonstrated the muscle activation was elevated nearly 10% to 20% and knee joint moments increased approximately 100% when study participants went from a planned to unplanned athletic maneuver. A more generalized co-contraction pattern was found in the muscle activations for the unanticipated maneuvers whereas a more selective pattern was utilized for the preplanned activities, suggesting the amount of time an athlete has to make a decision could alter the preparedness for the maneuver and altered biomechanical and neuromuscular strategies employed by the subject (Besier et al., 2003).

McLean and colleagues (2004a, 2004b) also performed a study utilizing two different conditions, with and without a simulated defensive player and determined the simulated defensive player trials resulted in dramatic increases in peak medial ground reaction forces, increases in hip flexion, hip abduction, knee flexion and knee valgus angles. The researchers performed these tasks with both male and female participants but revealed no statistically significant results for the impact of gender; however differences were noted including variability in hip rotation during the stance phase of the side step for males and knee rotation variability for females (McLean et al., 2004a; McLean et al., 2004b).

In a similar study, Ford and colleagues (2005) had subjects perform a jump-stop, and unanticipated cut maneuver to examine the knee flexion-extension angle as measured by a motion analysis system (Ford et al., 2005). This study was one of the very few to employ adolescent male and female athletes. Prior to each jump, the athlete was required to flex the knee to approximately 45 degrees and hold for four (4) seconds prior to the jump. Three jump directions were utilized for the unanticipated cut maneuver. The research group discovered females exhibited increased knee valgus (abduction) angles compared with males. Differences were also noted for the maximum ankle eversion and inversion during the stance phase between the genders. The researchers concluded that dynamic neuromuscular training for the adolescent athlete with an emphasis on frontal plane motion could help prevent ACL injuries (Ford et al., 2005).

While male and female sport participation should look very similar, it is apparent that differences exist in the manner of execution patterns between the genders. The athletic maneuvers performed by athletes are determined by several factors including maturation, experience level, leg dominance and fatigue.

Factors Affecting Athletic Maneuvers

Maturation

As athletes age, they gain more experience and see more scenarios. Maturational groups appear to affect the lower extremity alignment, and the development of mature alignment occurs in males and females at different rates (Shultz et al., 2008). According to Shea et al. (2004) differences between the genders in the incident rates for ACL injuries appear to begin to occur in athletes around age 12; however, the maximum number of female ACL injuries appears to occur later around age 16. It appears the injury ratio increases with the onset of puberty; however, the actual final injury risk might not stop climbing until middle to late puberty and into early adult development (Shea et al., 2004). Whatever mechanical movement strategy prepubescent children are employing for athletic maneuvers, it does not appear to exacerbate ACL injury risk, or the body is able to withstand the forces to which it is exposed (Hass et al., 2003). The identification of a child's landing strategy might provide adults with a model to adopt a "safer" landing strategy similar to the preparatory muscle activity employed by younger children to decrease the risk of injury (Hass et al., 2003).

Up until puberty, both males and females demonstrate similar neuromuscular control strategies during landing (Hewett et al., 2004). It appears neuromuscular control strategies change for females after the onset of puberty, and results in changes such as increased knee valgus alignment during landing sequences (Ford et al., 2003). Russell and colleagues (2007) have suggested a developmental difference between children and adults performing landing maneuvers, with the children relying on a strategy using the larger, stronger muscles of the hip and torso rather than the knee or ankle to control forces during the task. Adults appear to use more muscle activation from the hamstrings relative to the quadriceps (especially the vastus medialis), compared to children.

Experience Level

McLean and scholars (2005) suggest, "[M]ovement variability is largely dependent on skill/experience level" (McLean, Huang, & van den Bogert, 2005, p. 420). Novice athletes are possibly less adaptable to the physical demands of athletic activities (Traina & Bromber, 1997). Sigward & Powers (2006) suggested novice athletes could be at an increased risk of sports injury due to a variation in the kinematic pattern of the knee during side-step cutting maneuvers. The authors studied soccer athletes with varying levels of playing experience and discovered females with more soccer experience had larger knee moments, when performing athletic activities (Sigward & Powers, 2006). Few studies to date have examined the experience level of athletes and the risk of ACL injury between male and female athletes.

A study by Sabick et al. (2008) examined youth soccer players as they were dropped from a bar onto a force plate and then either run forward, side step cut 30°, or cross-over cut 30°. The study found significant differences between the genders during peak ground reaction forces in the center run and side cut. The study also suggested males tended to land with greater hip abduction, knee varus, and ankle inversion angles.

The novel approach of this study centered on the lack of data on unanticipated cutting tasks in adolescent (middle school) athletes (Sabick, 2008).

An additional study by Pfeiffer and colleagues (2007) examined the muscle activation differences between genders in the lower extremity. Youth soccer athletes participated in a drop land from a horizontal bar 30.5 cm above ground, and then were asked to cut 30° to the left or right. The group found no statistical differences for values between genders during landing. The vastus medialis did appear to have some significance during push off, with the boys reporting higher activation levels than the female participants. The authors indicated that males appeared to be "quadriceps dominant" in their landing strategy, which is usually associated with female athletes.

Leg Dominance

Limb dominance has been defined as increased dynamic control observed in one extremity over another as a result from imbalances in muscular strength or muscular recruitment patterns (Ford et al., 2003; Hewett, 2000; Knapik, Bauman, Jones, Harris & Vaughan, 1991). Strength imbalances in the lower extremity have been implicated as possible risk factors for lower extremity injury. Studies have demonstrated the differences in kinematics of the dominant and non-dominant lower extremity during athletic tasks (Nadler et al., 2002). Herman et al. (2008) suggested the muscular imbalance between the lower extremities could increase the risk for their extremity.

In a study by Nadler and colleagues (2002), athletes were more likely to injure the left lower extremity rather than the right. Nearly 90% of the time, athletes who are right leg (and right hand) dominant use the left leg for postural support during performance of athletic maneuvers (such as kicking a ball, or cutting to the right or running straight ahead) (Beling, Wolfe, Allen & Boyle, 1998). Overreliance on the dominant extremity can increase the stress on the joints, and further disrupt the muscular balance between the sides (Jacobs & Mattacola, 2005). The decreased strength in the non-dominant extremity can decrease the ability to absorb increased forces, such as those associated with athletic activities (Hewett, 2000). The dominant extremity of late or post-pubertal girls was significantly higher than the non-dominant extremity for knee valgus angles (Hewett et al., 2005).

Differences in the frontal and transverse plane have been noted for the lower extremity dominance during functional athletic activities such as landing from a jump (Ford et al., 2003; Jacobs & Mattacola, 2004). The differences in frontal plane kinematics are especially important to researchers, as the potential for increased knee ligament injury may occur with increased valgus motion. Authors have indicated leg dominance and strength differentials at the hip (i.e., the primary focus being the abductors) could be contributing factors to the reported functional differences (Jacobs et al., 2005).

Fatigue

The cause and definition of fatigue remains vague despite many years of direct research and study. When fatigue is applied to exercise, most researchers typically describe a general sensation of tiredness accompanied with a functional decrease in muscular performance capacity (Bilcheck & Kraemer, 1992; Chappell et al., 2002; Rowe et al., 1999). Bilcheck and Kreamer (1992) defined muscular fatigue as “a failure to maintain a required or expected force" (p. 9). Fatigue can also be classified as a failure of normal physiological functions such as when reductions of maximal force generating capacities are reached (Bilcheck & Kraemer, 1992). Most athletes have difficulty defining fatigue, but instinctively know when it occurs. When asked to describe fatigue, some athletes described a sensation of overwhelming tiredness and a functional limit to athletic participation (Griffin et al., 2000). Recently researchers have indicated the fatiguing process is initiated in conjunction with the start of athletic performance (Green, 1997; Griffin et al., 2000; Rowe et al., 1999). The researchers indicate each athlete has an energy "tank" which begins athletic participation at given levels, with greater initial tank energy stores staving off fatigue for longer durations. As energy is depleted during physical exertion, the energy stores of the body are used resulting in some level of fatigue and as the stores are exhausted, more fatigue is induced and performance declines (Griffin et al., 2000).

Muscular fatigue is one of the most significant contributing factors causing decrements and impairment in athletic performance. Physical conditioning programs are developed and implemented to delay the effects of muscular fatigue and enable the athlete to develop an increased functional capacity for competition (Bilcheck & Kraemer, 1992; Green, 1997; Griffin et al., 2000). Previous studies have suggested poor physical conditioning is associated with an increased rate of injury and the studies have indicated improved conditioning could reduce the probability of injury in athletes (Chappell et al., 2002; Toth & Cordasco, 2001). The association between fatigue and lower extremity injury, including ACL ruptures, remains anecdotal (Chappell et al., 2002).

The factors promoting fatigue are varied to the specific type of muscle action, mode and duration of the muscle action required for athletic participation (Bilcheck & Kraemer, 1992). The measurement of fatigue is quantified fundamentally by the measurement of force loss occurring in alliance with specific concentric, eccentric, or isometric muscle contractions (Green, 1997). However, this measurement does not account for the deprival of force due to the inactivation of muscle by the neural components (central) or inactivation of the excitation-contraction process within the muscle (peripheral) components (Bilcheck & Kraemer, 1992; Green, 1997).

Peripheral fatigue and central fatigue have been described as different phenomena, and either could influence athletic performance (Bilcheck & Kraemer, 1992). Central fatigue is described as theoretical impairment of motor pathways at voluntary motor centers arising from different impulses and receptors within the muscle (Madigan & Pidcoe, 2003). Peripheral fatigue is delineated by alterations within the fibers of skeletal muscles (Madigan & Pidcoe, 2003). A muscle's potential to resist fatigue is determined by the excitation and/or activation mechanisms housed within skeletal muscle fibers (Bilchek & Kraemer, 1992). Researchers argue over mechanism of fatigue, and some suggest fatigue could exist in either central or peripheral fatigue origin (Bilcheck & Kraemer, 1992; Madigan & Pidcoe, 2003).

According to research, ACL injuries are more likely to occur in the later stages of athletic competition suggesting fatigue is a contributing variable in the mechanism of knee injury. Data indicate a positive correlation in gymnastics between the duration of an athletic practice and number of injuries sustained (Chappell et al., 2005). Similar conclusions were reached in another study examining NCAA wrestling injuries. The researchers determined wrestlers were more likely to be injured in the third period of competition, suggesting fatigue could be a salient factor in the injury mechanism (Rowe et al., 1999).

During muscle contractions, force development is dependent on the number of attached actin and myosin crossbridges located within the sarcomere of the muscle (Green, 1997). Fatigue could result from a decreased abundance of crossbridge interactions or possible structural damage to the sarcomere arrangement (Bilcheck & Kraemer, 1992). During locomotion, fatigue has been correlated with decreased knee proprioception and increased joint laxity values (Green, 1997; Rowe et al., 1999). In addition, fatigue decreases the capacity of muscle fibers to absorb energy, and at the knee, altered neuromuscular function associated with fatigue has been determined to increase anterior tibial translational movements (Rowe et al., 1999).

Ground reaction forces, lower extremity kinematics, and muscle activation patterns during running, rapid stop tasks, and crosscutting have been studied to investigate the impact of fatigue. Late onset of quadriceps and hamstring muscle activation has been noted during running and rapid stop tasks. In addition, researchers noted premature knee flexion when fatigue is induced (Chappell et al., 2005). These altered neuromuscular and biomechanical changes are believed to decrease the shock absorption and knee stabilization capacity during landing (Toth & Cordasco, 2001). Researchers have also studied crosscutting tasks (in which the athlete is required to step across the midline of the body to make a cut), and determined quadriceps fatigue resulted in increased ankle dorsiflexion moments, decreased peak posterior breaking forces, decreased peak extension moments, and delayed peak knee flexion angles (Chappell et al., 2005; Rowe et al., 1999).

Fatigue of the musculature surrounding the knee presumably occurs during frequent and repetitive running and jumping tasks. Theoretically, the hamstring muscles could become fatigued resulting in a muscle that is no longer able to produce sufficient tension to reduce the anteriorly directed shearing force of the quadriceps muscle during athletic maneuvers, and subsequently no longer protects the knee against ligamentous injury (Fagenbaum & Darling, 2003). Fatigue of the hamstring muscle can result in decreased peak impact knee flexion moments, increased tibial internal rotation, and decreased peak ankle dorsiflexion during physical activity (Chappell et al., 2005; Toth & Cordasco, 2001).

Isokinetic exercises resulting in neuromuscular fatigue of the quadriceps femoris and hamstrings were analogous with an increase in anterior tibial translation; suggesting muscle fatigue could diminish dynamic knee stability (Wojtys & Huston, 1994). Dynamic knee joint stability relies heavily on neuromuscular control of the surrounding muscles (Rowe et al., 1999; Wikstrom et al., 2004). Hamstring muscle activation is provoked via stress placed on the ACL, which could alter the thigh muscle activation threshold and results in improved dynamic knee stability (Rowe et al., 1999). In opposition, the quadriceps has been implicated for their role in pulling the tibia anteriorly on the femur and stressing the ACL at knee angles close to full extension (Rowe et al., 1999).

Another gender study utilized an isokinetic fatigue protocol subjecting the quadriceps and hamstrings muscles to fatigue to analyze the anterior tibial translation, muscle reaction time, and muscle recruitment patterns of both genders (Wojtys et al., 1996). After the implementation of the fatigue protocol, the researchers found no difference when examining the order of muscular recruitment for both genders in response to forced anterior tibial translation. However, the amount of tibial translation was 32.5% greater in the fatigued state than in the non-fatigued state for subjects - indicating a potential slowing of the hamstring and quadriceps muscles in direct response to fatigue. The researchers also found the gastrocnemius muscle demonstrated a significant decrease in muscular response time even though it was not directly involved in the fatiguing protocol. The conjectured fatigue could alter the dynamic stability of the knee joint and could play an integral role in injuries sustained during physically demanding athletic maneuvers.

Athletic activity typically consists of prolonged activity that usually results in fatigue. A study conducted by Wojtys & Huston (1994) suggested fatigue could be a primary factor in the increased anterior tibial translational forces placed on the athletic knee during locomotion. Fatigue of the lower extremity muscles during physical activity may retard the potential dynamic stabilization and resulting knee defense mechanisms. The muscle firing in each of the medial and lateral quadriceps muscles was diminished by approximately 40% after fatiguing exercises were performed. The firing rate for the hamstring muscle saw similar decreases (lateral hamstring, 30% reduction in firing; medial hamstring, 35% reduction in firing) during fatigue exercises. This research indicates the muscle firing capability is greatly attenuated with the introduction of fatigue and could suggest a possible depreciation of muscular stabilization resulting from fatigue could allow for an increased anterior tibial translation possibly placing the ACL at risk for injury (Wojtys & Huston, 1994).

Fatigue has been examined to determine the kinetics and kinematics in stop and jumping tasks. Fatigue resulted in significantly increased peak anterior shear forces on the proximal tibia of recreational athletes, especially the female subjects (Chappell et al., 2005). An increase in anterior shear force could possibly increase the strain on the ACL and thus increase the potential risk of injury. The proximal anterior tibial shear force is increased with valgus moments at the knee and decreased with knee flexion angles. It appears male and female recreational athletes perform athletic maneuvers utilizing different lower extremity motor control strategies when performing a 3 stop-jump task. Differences in the lower extremity protocols could significantly contribute to the increased risk of knee injuries in the female athlete (Chappell et al., 2005). Even with a plethora of literature, researchers have yet to uncover the mystery surrounding fatigue and how it contributes to the noncontact ACL injury mechanism (Chappell et al., 2005).

Practical, Clinical, and Educational Implications

This study could provide vital information regarding injury rates for male and female basketball athletes and could contribute to the vast body of literature attempting to explain the injury rate phenomenon. The biomechanical evaluation of movement between the genders could provide essential information to coaches, parents, athletes, physicians, researchers and athletic trainers. This study could provide a basis for the creation and implementation of prevention programs and screening tools to identify athletes at risk for ACL injury.

Many parents, general practice physicians, and coaches are uninformed or misinformed about the traumatic effects of dynamic athletic activity and the risk of ACL injury to female athletes. Through this study, I have had contact with parents and had the opportunity to discuss the prevalence of ACL injuries in young female athletes. I have had the opportunity to discuss some of the findings research has suggested as possible predisposition factors in the risk ratio between the genders. Taking information from the technical jargon of research journals and turning it into information parents and young athletes can understand helps inform and educate young athletes and their families about potential risk factors and possibly ways to prevent injuries.

As an athletic training educator, I am charged with teaching athletic training students how to attempt to prevent injuries before they occur through pre-participation physical examinations (PPEs), visual acuity, and attention to detail when watching players practice and play. Once an injury has occurred, the athletic trainer is required to identify, assess, and treat any injury including the traumatic non-contact ACL injury. Through this study, it is my hope that we can identify some muscle recruitment patterns which might allow physicians, researchers, and athletic trainers to evaluate female athletes and identify athletes who are at an increased risk of knee injury.

Up until recently, researchers have primarily focused on the knee to determine the risk of injury. Through this study, I am attempting to find anomalies with the hip musculature possibly contributing to the increased risk of injury in female athletes. If we can find some differences, we can call attention to this area for screening and assessment protocols for athletes. Health care professionals could overlook the area of the hip as the athlete may not complain of pain or discomfort at the hip, only at the knee. The body remains a connected kinetic chain, and it appears important to examine the link above the knee for implications in the lower link.

As well, very few researchers have examined bilateral comparisons of lower extremity EMG, as most studies have used all of the EMG leads to examine the dominant leg (usually the right leg). This study could determine if the EMG analysis varies between the dominant and nondominant extremity. This information could provide vital information and create another analysis technique for athletic trainers to determine the risk of injury.

EMG assessment of muscular contractions is a noninvasive examination which can determine if an athlete is contracting one muscle faster than another or at a greater rate of firing. As previous research has indicated, female athletes (mostly adults) are more likely to have a quadriceps dominant muscle-firing pattern. Through EMG analysis, athletic trainers could identify athletes who display this dominant quadriceps pattern and develop a weight lifting, ploymetric, and functional exercise protocol to enhance the strength of the hamstrings which ultimately could help decrease the strain of the ACL from anterior tibial displacement during locomotion and physical activity.

Kinematic analysis of joint angles could provide similar information for professionals to create protocols to teach children and adolescent athletes how to move in methods that decrease the gravitational and leverage forces sustained to the body during dynamic movements including jumping, landing, and cutting. Kinetic analysis evaluates the amount of forces a body sustains during locomotion. Through the assessment of this information, we can attempt to educate athletes about their movements and create protocols to teach athletes to land differently, sustaining a decrease in the forces, or learning to distribute the forces through the kinetic chain in the body.

In addition, it is extremely important to call attention to the adolescent athlete population. Athletic trainers are commonly employed in collegiate and high school settings; however, as of yet, most junior high schools and middle schools currently rely on coaches and parents to provide training/conditioning, prevention strategies, and medical treatment to their athletes. Children are engaging in athletic activities at extremely young ages, and are competing at elevated levels. They are subjected to injury during every practice or game. Research has suggested male and female athletes are injured at similar rates up until the onset of puberty, therefore, this study is integral in providing insight to educators and researchers to attempt to further the examination into the injury process. If this study provides unique information, then information could be used to attempt to educate athletic training students about the injuries sustained by younger athletes than they are typically trained to deal with in common professional settings. In addition, if we can determine differences between the genders, then we might be able to provide athletic trainers, physicians, coaches, and parents with the opportunity to assess risk of injury in these younger athletes to attempt to slow the rate of injury.

Although few people involved in athletics have access to complex machines such as an EMG or a VICON motion analysis unit, functional assessments can be made by the lay person and the research can be applied through the creation of specific training protocols, extending pre-participation physical examinations, education of athletes and their parents, and creating awareness among health care professionals such as athletic trainers and physicians. The risk of injury for the adolescent female athlete is great, the disability after injury is substantial, and therefore, anything to prevent injuries should be done by those in a position of control to stem the risk of injury.

Summary

In conclusion it is unclear exactly what mechanism causes noncontact ACL injuries and the risk factors associated with the increase in incidence in injury although much research has been dedicated to finding answers (*Figure 2.3*). The previous research has attempted to explore all of the options and add to the body of knowledge; however, the influence of the hip in the kinetic chain of the lower extremity has largely remained under explored. Researchers have failed to determine why prepubertal children appear to sustain injuries at similar rates until the onset of puberty at which time injury rates appear to spike. It is unclear why female athletes progressing through puberty end up with injury rates of anywhere from two to eight times higher than their male counterparts. Male and female athletes participate in sports at increasingly young ages at competitive levels rivaling older participants. Neuromuscular differences between the genders appear to be the most prominent and modifiable factors in the increased rate of injury. It appears differences in muscular activation levels, joint kinematics, and ground reaction forces between the genders after the onset of puberty could play an essential role in determining the risk of injury. This study will attempt to assess the four major muscles surrounding the hip and determine if differences exist based on gender for adolescent athletes during unanticipated cutting maneuvers. Identification of differences could allow professionals the opportunity to educate young athletes and their families about the risks and attempt to create training protocols to reduce the risk of injury.

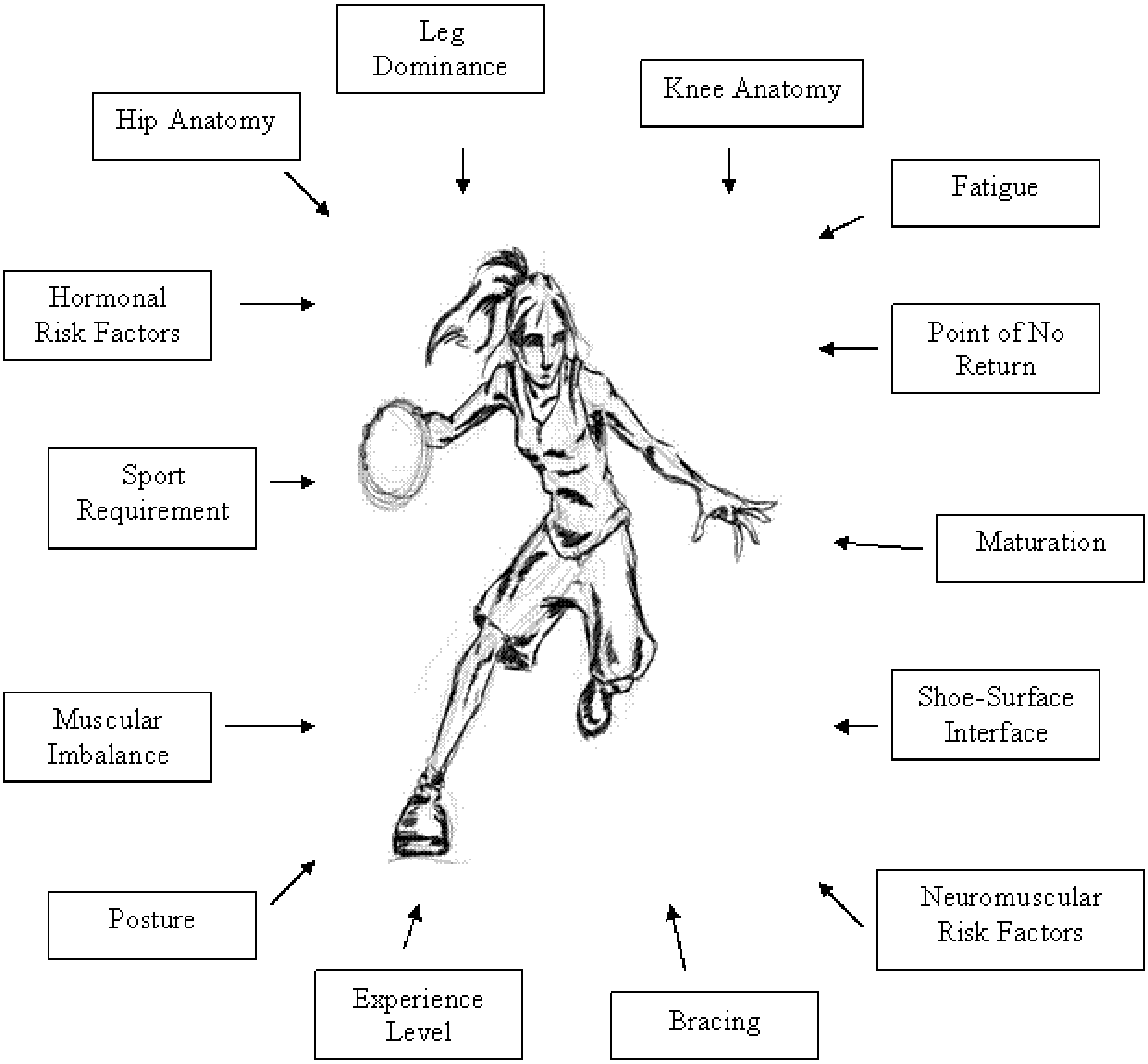


Figure 2.3. Risk Factors Affecting The Female Athlete During Athletic Participation.

CHAPTER 3

Research Methodology

The purpose of this study was to determine the effects of gender on a jump, land, and unanticipated cut in adolescent male and female basketball athletes. Previous research has suggested different muscle activation patterns are utilized by each gender when performing specific types of athletic maneuvers. In addition, literature suggests women display less knee, hip and trunk flexion during gait and landing tasks compared to males (Decker et al., 2003; DiStefano et al., 2005; McLean et al, 2004b; Salci et al., 2004; Yu, Lin, & Garrett, 2006). These studies have suggested sagittal plane coupling of the hip and knee could be determining factors for the risk of ACL injury. These differences have been documented not only at the knee, but recent research has examined the effects occurring at the hip as well. This chapter will discuss the (a) subjects, (b) instruments and apparatus, (c) procedures, and (d) design and analysis used when comparing the effects of gender on athletic activity.

Subjects

Ten adolescent basketball athletes (Males: N = 5 Females: N = 5) between the ages of 13-17 were recruited from local treasure valley club teams. Participants were required to have participated in the sport for a minimum of 1 or 2 years of playing experience at the club level (Table 1).

Table 3.1

Participant Demographics

	Males	**Females**
Subjects	N = 5	N = 5
Age (years)	15.2 ± 1.9	14.6 ± 1.3
Height (cm)	181.6 ± 3.6	177.8 ± 8.0
Weight (kg)	72.7 ± 6.0	65.9 ± 7.1
BMI	20.5 ± 1.2	20.8 ± 2.1
Maximum Vertical Jump (cm)	54.0 ± 8.2	41.9 ± 4.3
Years Playing Basketball	5.3 ± 2.9	7.0 ± 1.9
Years Playing Club	2.0 ± 1.3	2.0 ± 0.8

Arthropometric measurements (height, weight, age [date of birth], years of playing experience and dominant limb of the subjects) were obtained from the participants. The dominant limb was self reported by the participant by asking the participant which leg they would prefer to use to kick a soccer ball. All five of the male participants reported the right leg as their dominant limb, and four of the five female participants reported the right leg as their dominant limb. The participants were asked to wear specific types of clothing to gather proper data. Participants were asked to wear tight fitting clothing, but were allowed to wear baggy shorts, which were tucked into the bottom of spandex worn under the baggy shorts.

An independent samples t-test was run on the demographic data to determine if significant differences were noted between the genders. Height (p = 0.013), weight (p = 0.005), and vertical jump (p = 0.006) were all statistically significant demographic values. The male participants weighted more, were taller, and jumped higher than the female participants in this study. None of the other demographics were significant.

All testing was performed at Boise State University Biomechanics laboratory. The Institutional Review Board at Boise State University approved this study prior to initial correspondence with the coaches, parents, and athletes; approval # BM 103-09-002 *(Appendix A).*

Participant's Training Regiments

Males

Two participants did not participate in any training regiments other than their team basketball practices. Two participants reported engaging in weight training session approximately 2-3 times per week. Three participants reported participating in sports other than basketball including: football, track, soccer, and baseball.

Females

All five participants were involved in training in addition to their regular basketball regiment. One participant was involved in weight training approximately 3 times a week. One participant was involved in agility training approximately 3 times a week. Three participants were involved in sports other than basketball including: soccer, volleyball and running. One participant was involved in a core-strengthening program every day of the week.

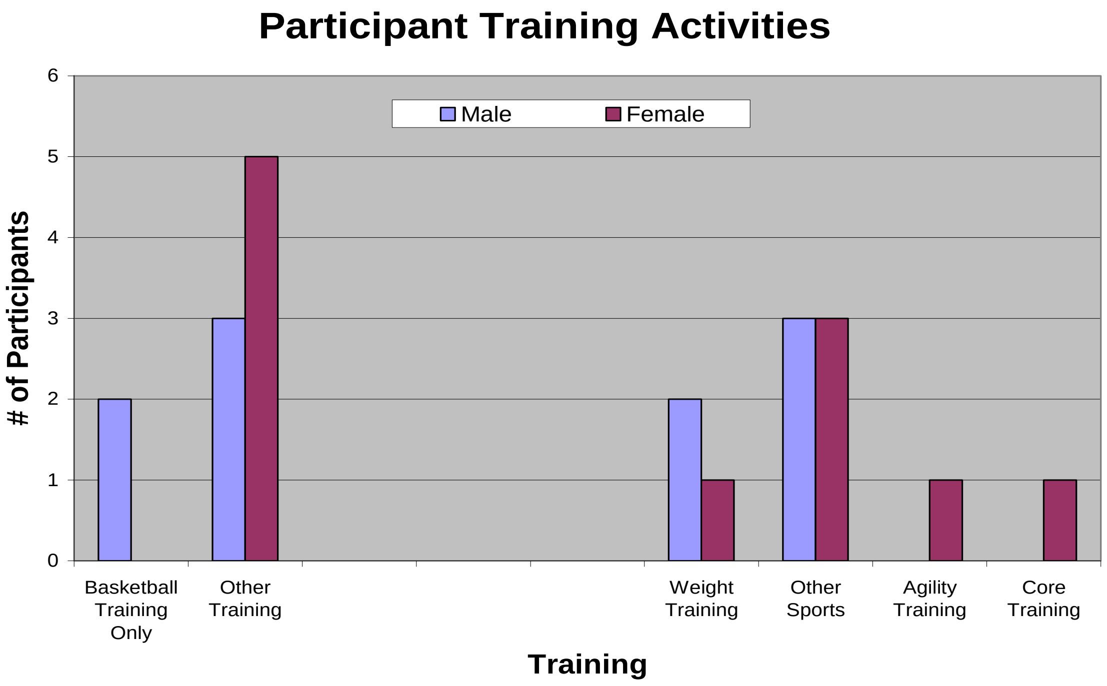


Figure 3.1. Distribution Of Training Protocols Among Study Participants.

<u>Instruments and Apparatus</u>

Data collection occurred during a one-time data collection session lasting approximately 1½ - 2 hours. The session was comprised of seven parts: Subject preparation; Motion Analysis; EMG Assessment; Warm-up; Isokinetic Assessment; Maximum Vertical Jump Height Calculation; and Jump, Land, and Unanticipated Cut Assessment. After the participant and his or her parent or legal guardian was briefed on the scope of the study, read the "Research Participation Bill of Rights," asked any remaining questions, and signed the informed consent form (parent or legal guardian) (*Appendix B*) and informed assent form (participant) (*Appendix B*), and then the participant was prepared for the study. Immediately following the preparation the participant was asked to complete a warm-up session (*Appendix C*). Following the warm up, the participant was tested for his/her maximum vertical leap. The participant was then introduced and tested using the jump, land and unanticipated cut protocol while researchers collect data via motion analysis and EMG assessments. After the functional testing protocol, an isokinetic assessment tested the participants for maximum voluntary isometric contractions (MVIC's) for hip flexion, extension, abduction, and adduction using the isokinetic dynamometer.

Vertec® Vertical Jump Assessment

The Vertec® Vertical Jump assessment was used to calculate the participant's vertical leap. The participant was required to stand up as tall as possible and reach each hand overhead until the hands cross over each other. With the feet still on the ground (no tippy toes), the participant was asked to walk under the Vertec® poll and move as many marker bars as possible to get a baseline value. The participant was then required to perform a standing vertical jump. No step approach was allowed, and the participant was required to stand directly under the Vertec® poll and squat down and jump as high as possible to move the marker bars. The Vertec® measures jump height in half-inch increment markers therefore maximum vertical leap was calculated in half-inch markers.

The male participants generally performed maximum vertical leaps that were higher than their female counterparts. On average, males had maximum vertical jumps that were 54.0 cm ± 8.2 cm, whereas the female participant's maximum vertical jumps averaged 41.9 cm ± 4.3 cm.

Ground Reaction Forces

Ground reaction force (Kistler, Type 9821C) was collected with two in-ground force plates in the BSU Center for Orthopaedic and Biomechanic Research (COBR) lab. The force plates were sampled at 1000 Hz. The force plates measured: peak anterior/posterior ground reaction forces during landing and push-off; peak medial/lateral ground reaction forces during landing and push-off; and peak vertical ground reaction force during landing and push-off. The force plates were oriented so each leg (right and left) has a global coordinate system. The force plates were oriented so the X-axis force values coming from the lateral left foot or lateral right foot are positive in force value, and forces from the medial left foot or medial right foot are negative. The Y-axis reported anterior forces as positive values and posterior forces as negative value. The Z-axis recorded vertical forces. All force values were normalized for subject bodyweight in order account for body mass when comparing between subjects (*Appendix D*).

VICON Motion Analysis System

Motion Analysis assessments were captured via a VICON ® motion capturing system consisting of 7 stationary infrared cameras scattered throughout the laboratory room. Each camera was calibrated with motion analysis equipment and then again with a static trial for each participant. If calibration values fell above a desired range, then the calibration was performed again until adequate levels were reached.

Biodex Isokinetic Machine

The Biodex Isokinetic Dynamometer® was used to assess Maximum Voluntary Isometric Contractions (MVIC). The MVICs were collected to normalize the electromyography (EMG) signals during analysis and thus allow for comparison between and among subjects. The Biodex was used as a static force to collect EMG data from an isometric contraction for both the right and left leg in the four motions of the hip (flexion, extension, abduction, and adduction). The lever arm kept constantly at 135º for all participants across all trials (both right and left for the four muscles). The participants were instructed to maximally contract against the lever arm, hold for a count of three seconds and then relax (*Appendix E*).

Procedures

Each participant was required to report to the BSU COBR lab to complete a one-time data collection. The minor participants read and signed an informed assent prior to data collection (*Appendix B*). Each participant was required to bring a parent or guardian to the testing session and the adult was asked to read and sign an informed consent to allow their son or daughter to participate in the study (*Appendix B*). Each participant was instructed to start standing on a pressure sensor. Upon verbal cues from the researcher, the participant was to jump over a barrier, land with one foot on each of the inground forceplates, and side cut in a specific direction. As the participant jumped over the barrier, the pressure sensor would signal a light for one of the three cutting directions to appear on a board while the participant was still in flight. Upon landing, the participants were instructed to make a side step (right light indicated the participant should cut to the right leading with the right leg and using the left leg as a plant leg) and immediately run towards cones approximately 4.6 m (15 feet) away. The three cutting directions were: 30° degrees to the right, straight ahead, or 30° degrees to the left. A minimum of fifteen (15) randomized jump, land, and unanticipated cuts were performed so each subject had 5 good trials in each of the three directions (*Figure 3.2*).

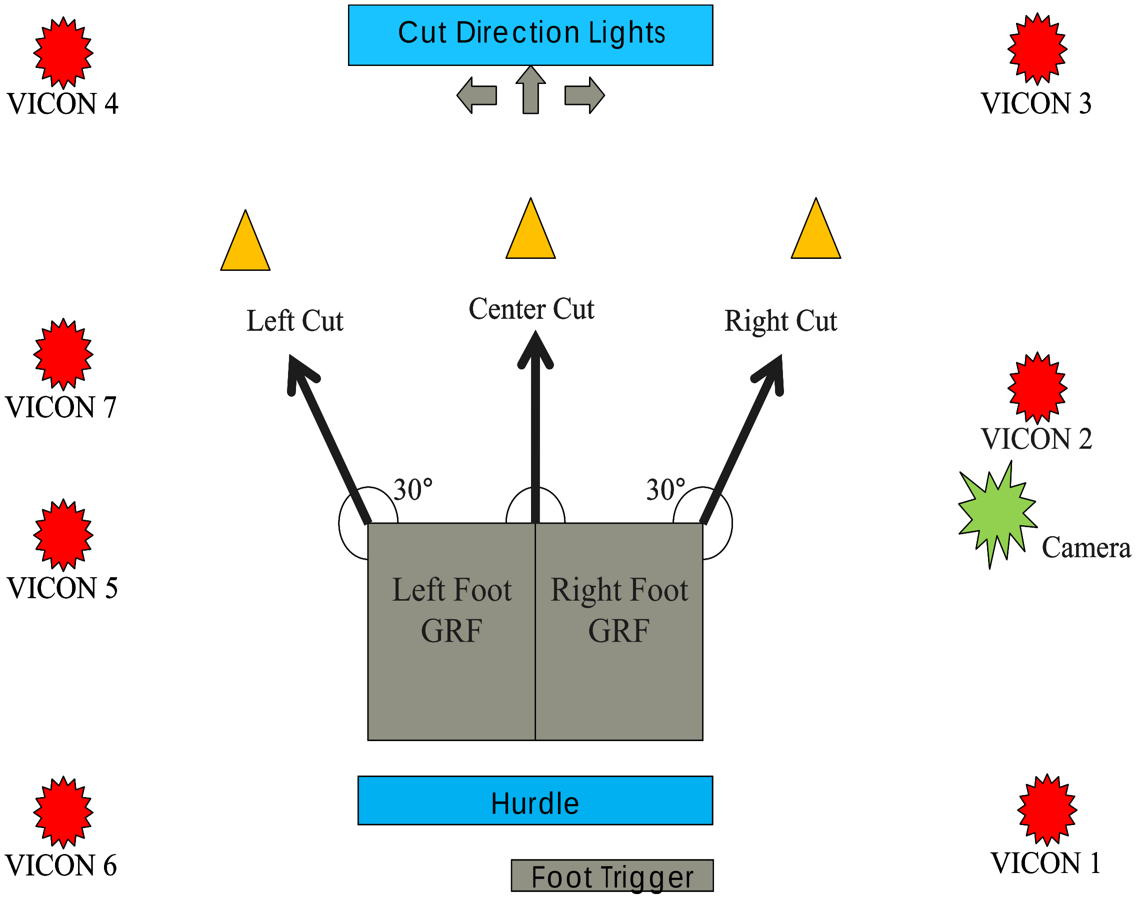


Figure 3.2. Schematic Of The Laboratory And Jump, Land, And Cut Task

In *Figure 3.2*, the participants are monitored by 8 cameras (7 VICON retroflective cameras, and 1 digital video camera) during testing. There are more VICON cameras placed on the participant's left side because there are a bank of windows within the laboratory which makes collecting data difficult. Any reflection from an outside light source can affect the cameras ability to "read" or "pick up" the markers as they move through space, so in an effort to collect more consistant data, the 7th VICON camera was placed on the left side. At any given time, at least two of the seven VICON cameras must be able to "see" a marker in order for it to be recorded. If the marker is "missing" or was not recorded, the marker could possibly be placed back into the sequence if the time gaps between marker sightings are small enough using the BodyBuilder Software (Vicon BodyBuilder, Version 3.6 – build 141, Oxford, United Kingdom). If too much time has eclipsed between marker sightings, the trial must be disguarded and not used for analysis.

Warm-Up

The warm-up was the same for all participants in the study. The warm-up was conducted in the biomechanics lab and consisted of light callisthenic activity such as jogging, and dynamic stretching (slow high knees, slow squats, grapevine, etc.), and some traditional speed warm-up drills (fast, low intensity skips and hops). The warm-up was concluded with a light stretching procedure to reduce the risk of injury during participation *(Appendix C).*

Subject Preparation

Anthropometric measurements (height, weight, age [date of birth], years of playing experience and dominant limb of the subjects) were obtained from the participants. The dominant limb was self reported by the participant by asking them which leg they would prefer to use to kick a soccer ball. Participants were asked to wear specific types of clothing to gather proper data.

The skin superficial to the muscle belly of right rectus femoris, left rectus femoris, right biceps femoris, left biceps femoris, right adductor longus, left adductor longus, right gluteus medius, and left gluteus medius was prepared by cleaning it with isopropyl alcohol. Researchers placed electrodes and reflective markers on the participant's hips and thighs, and therefore, requested specific athletic wear for the participants of the research study. Female participants were asked to wear a one-piece swimming suit or sports bra (tight fitting tank top) and spandex/compression shorts during data collection. Male participants were asked to wear spandex compression shorts (bicycle shorts) during data collection. Both male and female participants were allowed to wear their basketball shorts over the spandex and tuck the bottom of the shorts into the spandex during data collection for comfort.

Maximum Vertical Leap

The participants were asked to perform three (3) jumps to calculate a maximum vertical jump height. Three jumps were used to attempt to introduce the participant and then obtain their maximum vertical jump height. The researcher used a Vertec vertical jump analysis tool to calculate the average maximum vertical leap for an individual participant. The researcher calculated 75% of the maximum vertical jump height and placed a hurdle at that height in front of two ground reaction force plates. During a pilot study using similar subjects, researchers began with jump heights of 50%; however, when the data was analyzed, the research team discovered the male and female participants were jumping a much greater heights when clearing the hurdle. As such, the research team decided to make the task a little more difficult and increase the hurdle height to 75% of the maximum vertical jump in order to make the task more challenging.

EMG Assessment

The researcher prepared the participant's skin for placement of electromyography (EMG) electrodes to determine the actions of the muscles under the skin. An 8-channel surface Electromyography (EMG) system (BTS Free EMG, Italy) was used to collect electrical activity of the bilateral muscles surrounding the hip from small diameter (12 mm), round, silver/silver chloride, bipolar, preamplified electrodes (Myotronics, Inc., Kent, WA). EMG signals were sampled at 1000 Hz and interpreted by MYOLAB and processed with MATLAB software (The MathWorks Inc.). The peak mean amplitude was used to analyze muscle activity during the athletic maneuver. The peak mean amplitude was calculated by taking each trial and finding the maximum amplitude and then averaging that maximum over the number of trials (*Figure 3.3*).

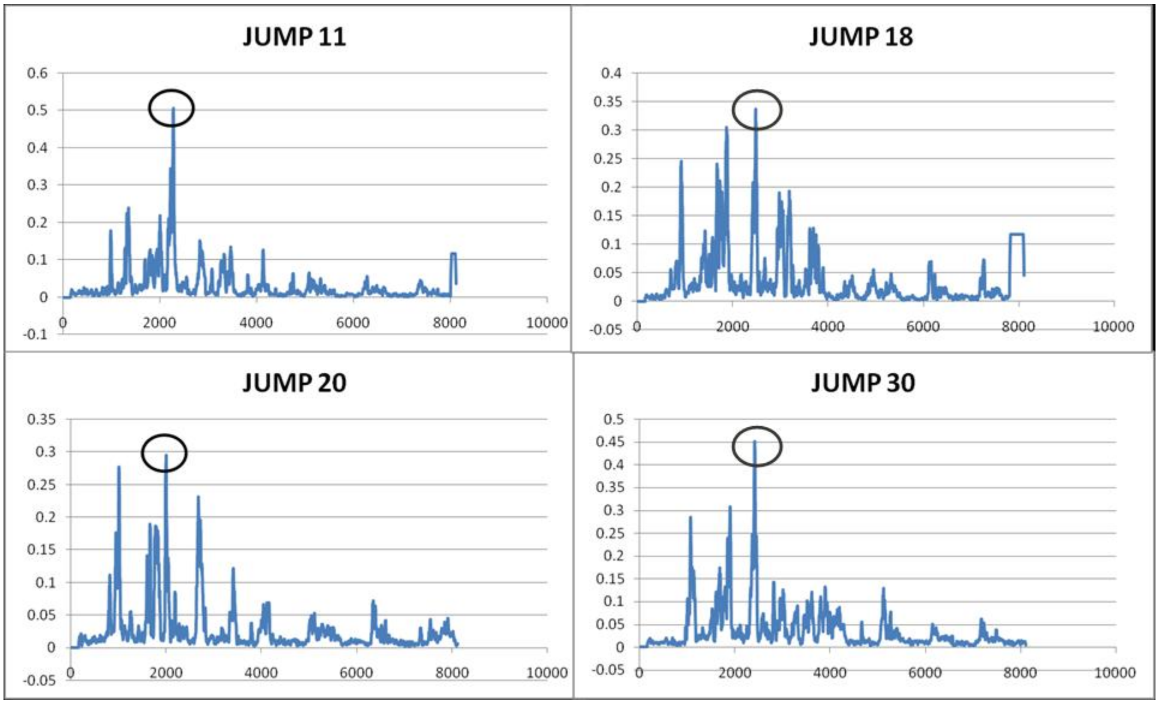


Figure 3.3. Four Left Cut Trials For A Male Participant With Circles Over Each Peak Amplitude Which Are Averaged To Calculate The Peak Mean EMG Amplitude.

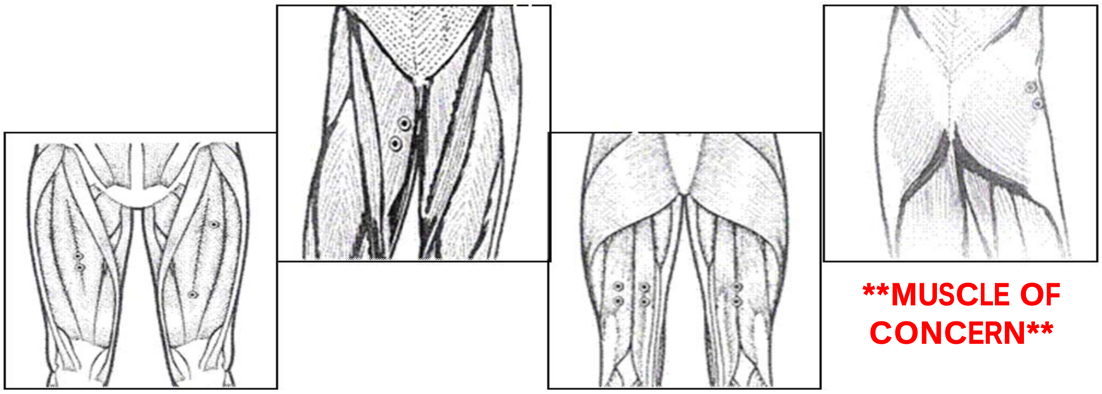
The skin was shaved to remove hair (if needed), then rubbed with an alcohol swab to remove excess dead skin and create good conduction for collection of EMG data. After proper skin preparation, a total of eight (8) EMG surface electrodes were placed on each participant's left and right hip region: 1) Gluteus Medius (Hip abductor); 2) Proximal Hamstrings (Hip extension); 3) Proximal Quadriceps (Hip flexion); and 4) Adductor Longus (Hip adduction). All EMG placements were determined by following the protocols listed in *Introduction to Surface Electromyography* (Cram, Kasman & Holtz, 1998). Participants were tested for isometric muscle strength on an isokinetic dynamometer to obtain a maximal voluntary isometric contraction (MVIC) for each of the muscle groups tested to be used as a normalization tool for the EMG analysis.

Gluteus Medius EMG Placement. The researcher palpated the iliac crest (hip bone) to locate the gluteus medius muscle just inferior to the bony landmark. Active electrodes were placed parallel to the muscle fibers over the proximal third of the distance between the iliac crest and the greater trochanter (*Figure 3.4*).

Hamstring EMG Placement. The researcher palpated the posterior (back) of the proximal thigh. The muscle is located on the center of the posterior surface of the thigh, approximately half the distance between the knee and the posterior iliac spine. Active electrodes were placed parallel to the muscle fibers over the muscle belly (*Figure 3.4*).

Quadriceps EMG Placement. The researcher palpated the rectus femoris muscle. The muscle is located on the center of the anterior surface of the thigh, approximately half the distance between the knee and the anterior iliac spine. Active electrodes were placed parallel to the muscle fibers over the belly of the muscle (*Figure 3.4*).

Adductor Longus EMG Placement. The researcher palpated the medial thigh while asking the participant to perform an isometric contraction for hip adduction. The muscle is located on the medial thigh just inferior to the pubic symphysis. Active electrodes were placed on the medial aspect of the thigh in an oblique direction approximately 4 cm from the pubis (*Figure 3.4*).



MUSCLE OF CONCERN

Photo: Cram, Kasman, & Holtz,1998

Figure 3.4. Electromyography Marker Placement (Left To Right: Rectus Femoris [Quadriceps], Adductor Longus, Biceps Femoris [Hamstrings], And Gluteus Medius).

Motion Analysis

The participant was fitted with reflective markers by the researcher in order to obtain information about the joints and body lever systems during movement. The reflective markers were attached to participant's body using double sided tape and/or elastic bands. Motion Analysis assessment was captured via a VICON motion capturing system consisting of 7 cameras scattered throughout the laboratory room. Approximately 32 reflective markers were used to create a total body image for three-dimensional analysis of athletic maneuvers. A total body marker system was used to create a whole skeletal system based on the VICON® plug-in gate marker set up (*Figure 3.5, Appendix F*).

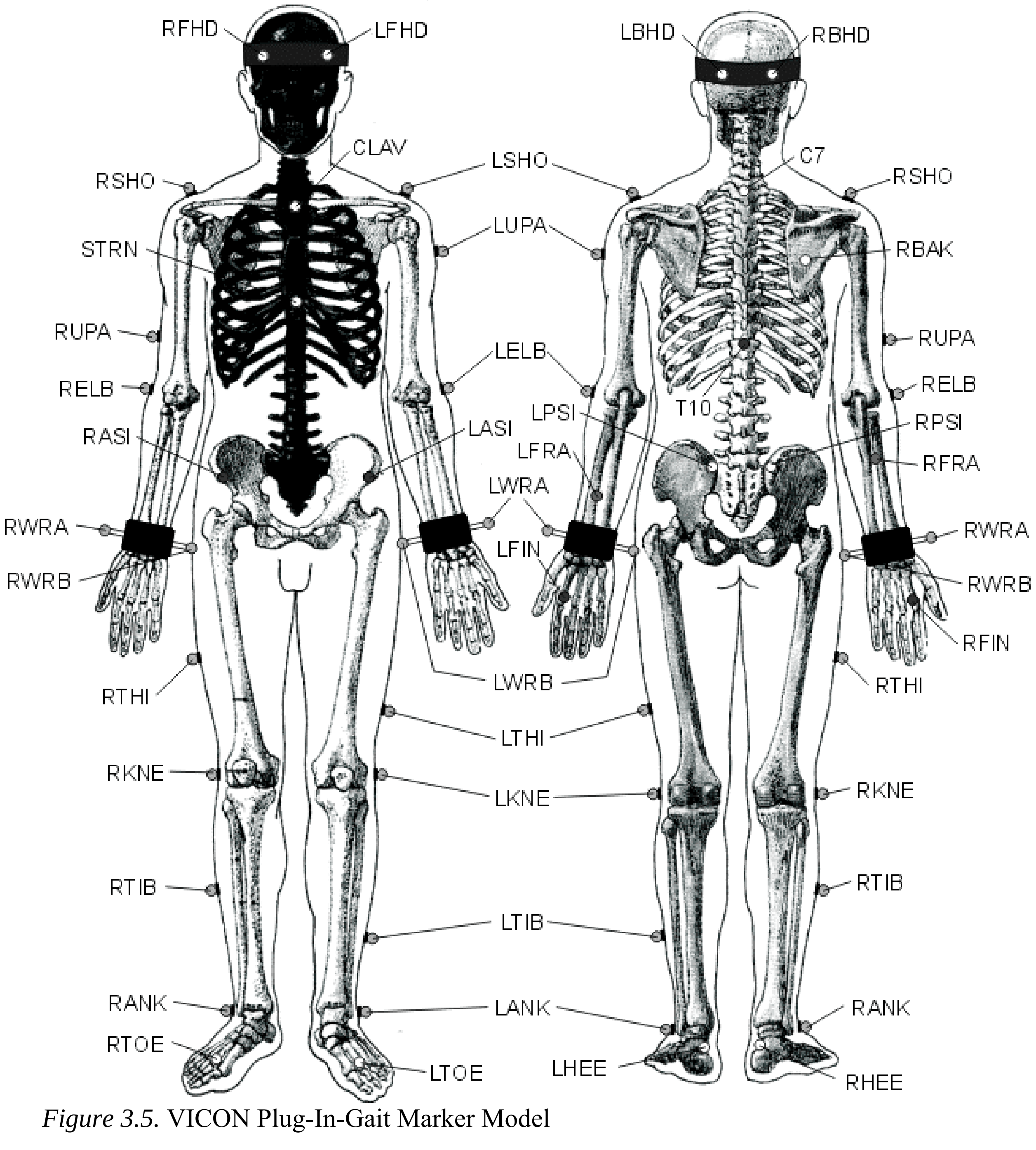


Figure 3.5. VICON Plug-In-Gait Marker Model

A VICON® (VICON Motion Systems, Lake Forest, CA) motion analysis system consists of 7 infrared cameras controlled by Nexus® software (VICON Motion Systems, Lake Forest, CA) provided joint position data during the jump, land and unanticipated cut maneuvers. Each camera was calibrated with motion analysis equipment prior to data collection and then again with a static trial for each participant. If calibration values fell above a desired range (0.2 pixel image error) then the calibration was performed again.

Jump, Land, and Unanticipated Cut Assessment

The participants were introduced to the jump, land and unanticipated cut maneuver. The athlete was given several practice attempts to master the skill of jumping over a barrier to land on a force plate. The participant was required to land with a single foot on each of the force plates or the trial was not counted as successful. The force plate jump distance was approximately 120-150 cm. The participant was shown a light to direct the cutting movement in one of three directions (30° to the right, straight ahead, 30° to the left). Upon landing, the participants were instructed to make a side step (right light indicates the participant cut to the right leading with the right leg and using the left leg as a plant leg) (*Figure 3.6*, *Figure 3.7*, & *Figure 3.8*). The athletes were tested with a minimum of fifteen jumps, five (5) jumps in each of the directions presented in a randomized fashion to gather data on all participants. After participants finished the athletic protocol, the electrodes and reflective markers were removed from the participant. The participant and his or her parent were debriefed and then dismissed from the research study.

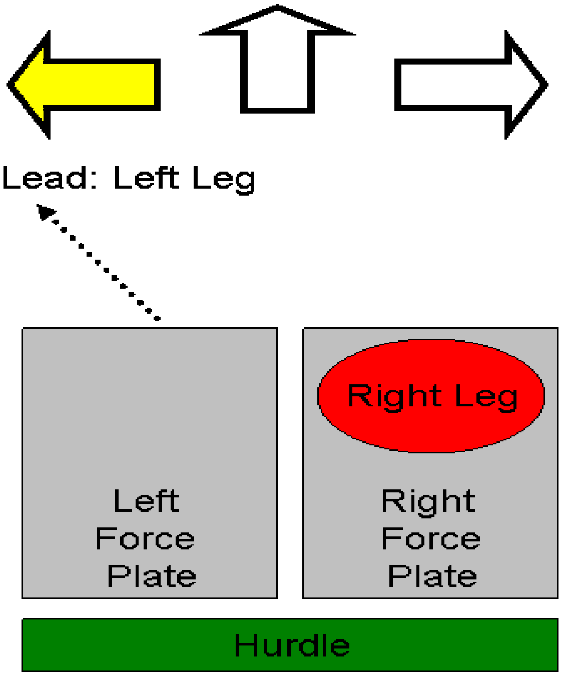


Figure 3.6. Left Cut: Participant Cuts Off Of The Right Leg (Dominant Leg) And Leads With The Left Leg (Non-Dominant Leg) To Initiate A Left Side Cut

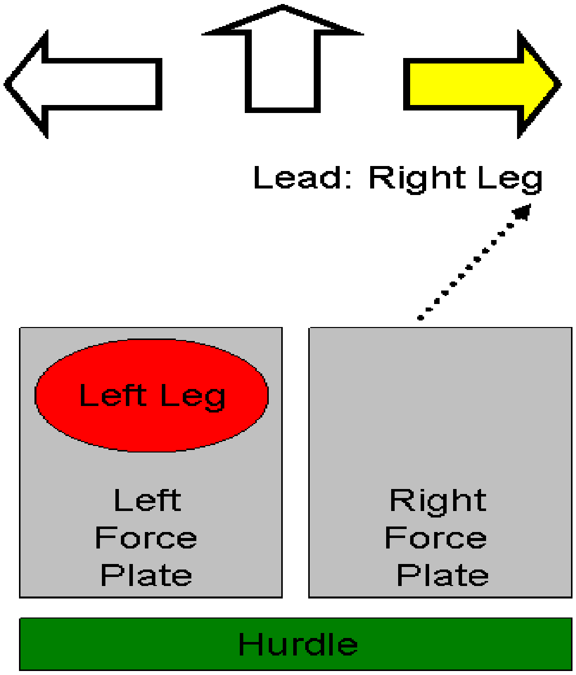


Figure 3.7. Right Cut: Participant Cuts Off Of The Non-Dominant Leg And Leads With The Dominant Leg To Initiate A Left Side Cut

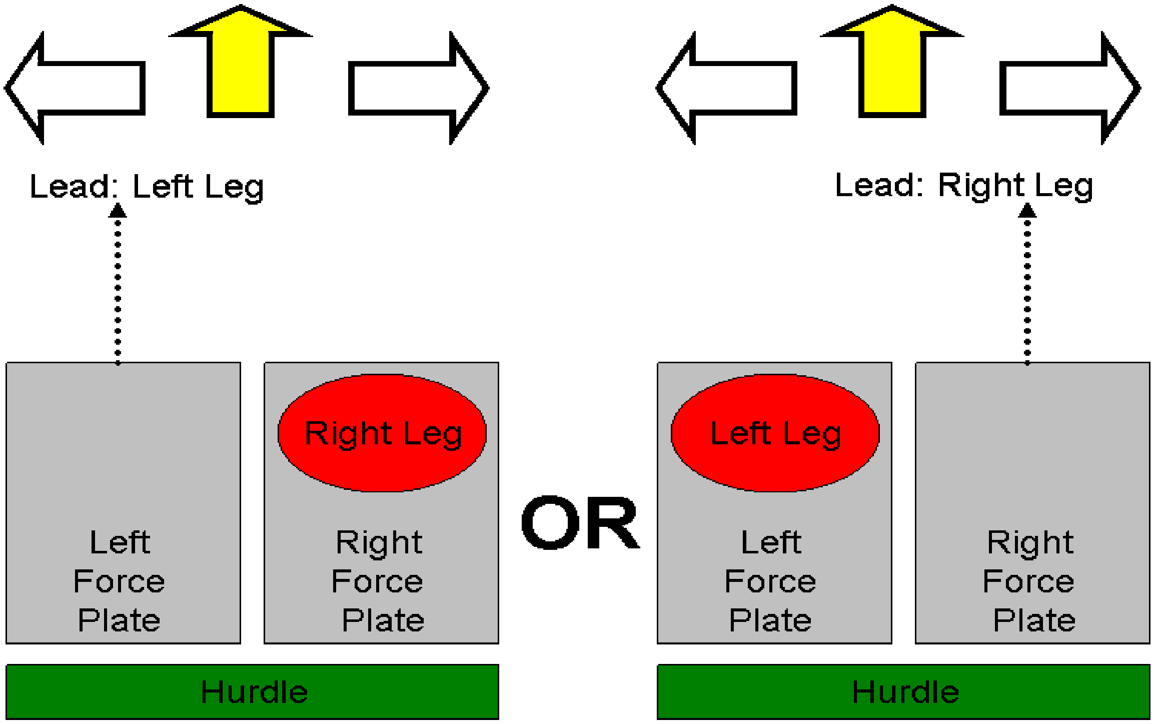


Figure 3.8. Center Cut: Participant Can Either Choose To Cut Off Of The Right Leg (Dominant Leg) and Lead With the Left Leg (Non-Dominant Leg) Or Can Cut Off Of The Left Leg (Non-Dominant Leg) And Leads With The Right Leg (Dominant Leg) To Initiate A Straight Run (Center Cut).

Two multiaxis ground level force plates (Kistler, Type 9821C) collected ground reaction forces. The force plates were sampled at 1000 Hz. Landing force was defined for each leg as the moment when the force plate detected any vertical component (Fz greater than 20 N) of a ground reaction. The force plates were oriented so the X-axis force values coming from the lateral left foot or lateral right foot are positive in force value, and forces from the medial left foot or medial right foot are negative. The Y-axis reported anterior forces as positive values and posterior forces as negative value. The Z-axis recorded vertical forces. All force values were normalized for subject bodyweight in order account for body mass when comparing between subjects *(Appendix D)*.

Isokinetic Assessment

The participants were fitted and tested on the Biodex System II® isokinetic dynamometer to measure the maximum voluntary isometric contraction of (MVIC) the four muscles of the left and right hips (gluteus medius, hamstrings, quadriceps, and adductor longus) to facilitate the electromyography (EMG) analysis. The participant was fitted on the Biodex II isokinetic machine to test hip strength. The participant was asked to lie supine on the isokinetic machine. A standard knee attachment device was secured to the proximal leg so that the pad was placed between the knee and the hip. The Biodex was used as a stationary force to collect EMG data from an isometric contraction for both the right and left leg in the four motions of the hip. The lever arm of the isokinetic dynamometer was held stationary at 135° during isometric testing of the hip for flexion, extension, abduction and adduction. The research participant was required to perform 3 repetitions of isometric contractions (contracting the muscles without moving the joint) to assess the maximum strength for the right and left hips during hip flexion, hip extension, hip adduction, and hip abduction.

Data Processing

Electromyography Signal Acquisition

Electromyography (EMG) signals cannot be seen by the naked eye so they must be located and transmitted by an amplifier to be able to be analyzed. The body filters electrical signals as they pass through different tissues. The EMG signal can also be filtered as it is amplified and recorded, therefore, it is important to examine the different processes an EMG signal is subjected to prior to being presented as usable research data.

Physiological EMG Signals

Physiological EMG signals are housed in the motor unit action potentials (MUAPs) of the muscle fibers. As a muscle depolarizes to initiate a contraction, an electrical impulse is sent through the muscle which causes the muscle to shorten. Physiological EMG signals cannot be measured as they emanate from the surface of muscle fibers. The muscle fibers are located deep in the muscle beneath layers of tissue. The electrical signal must travel through subcutaneous tissue (i.e., muscle, fascia, and adipose tissue) in order to be recognized by a surface electrode.

Tissue(s)

As electrical signals pass through tissues they encounter anisotropy which means they have a directionally dependent resistance to tissue based on the absorbance, refraction, and density of tissue. This anisotropy creates a low pass filter of the data. The greater the density of the subcutaneous tissue, the greater the spatial filtering to which the signal is subjected, which results in a greater low pass filter. Additional subcutaneous tissue can reduce the median value of the signal frequency. The fatty layer can create a type of insulator which decreases or stops the flow of electrical current. All of our subjects BMI values were within a relatively small range and therefore, it is assumed that the amount and distribution of body fat in each of the participants did not alter any of the

EMG signal collected during the study. The electrodes are placed on the skin, parallel to the muscle fibers to create a good conduction. The electrodes are placed slightly off center of the muscle belly where motor end plates are in the greatest concentration and the action potential given from the muscle can be recorded.

Electrode-Electrolyte Interface

EMG electrodes typically have a metallic portion in order to collect the electrical signal. The electrode-electrolyte interface occurs at the contact layer between the metallic detection surface of the EMG electrode and the superficial conductive tissues creating an electrochemical junction. This junction behaves as a high pass filter during EMG signal acquisition.

Bipolar Electrode Configuration

EMG electrodes can be presented in a monopolar or bipolar configuration. A monopolar electrode configuration consists of a single electrode and a reference electrode. A bipolar configuration (more common in current units) consists of two electrode-electrolyte interfaces or two signals collected from a single muscle which must be fed through a differential amplifier prior to analysis. In a bipolar configuration, two electrochemical junctions are located in near proximity to each other to monitor the same muscle, but each junction gets a slightly different signal from the underlying motor unit action potentials occurring during muscle contraction. Bipolar electrode configurations create a bandpass filter as the two signals are compared to each other during amplification. The distance between the two signals (interdetection surface spacing) of the bipolar electrode configuration should be very minimal. The interdetection surface spacing should be approximately 1.0 cm as a recommendation. When the interdetection surface spacing increases, the detection of EMG amplitudes from adjacent and deep muscles is increased thus creating the problem of cross-talk, where EMG signals are picked up from other muscles which are not originally intended.

Recorder

Each EMG recording device has different settings regarding the filtering method utilized when expressing raw EMG data. Sometimes the filtering paradigm is created by the EMG manufacturer and sometimes it can be created by the user. Some units have modifiable filters and others do not. The EMG unit used for this study, BTS FreeEMG, has a high pass filter programmed into the hardwiring of the collection unit.

Observable EMG Signal

Once the electrical signal has passed through all of the previous filtering structures, it is presentable as an EMG signal which we are use to seeing. The raw EMG signal is stochastic in nature and appears on both the positive and negative side of the horizontal axis. Raw EMG signal gives researchers a lot of information, as it can indicate when muscle burst is occurring and gives us an idea of the strength of the contraction; however, it is presented in an unusable form. The data must be transposed prior to analysis in order for it to be comparable between and within participants.

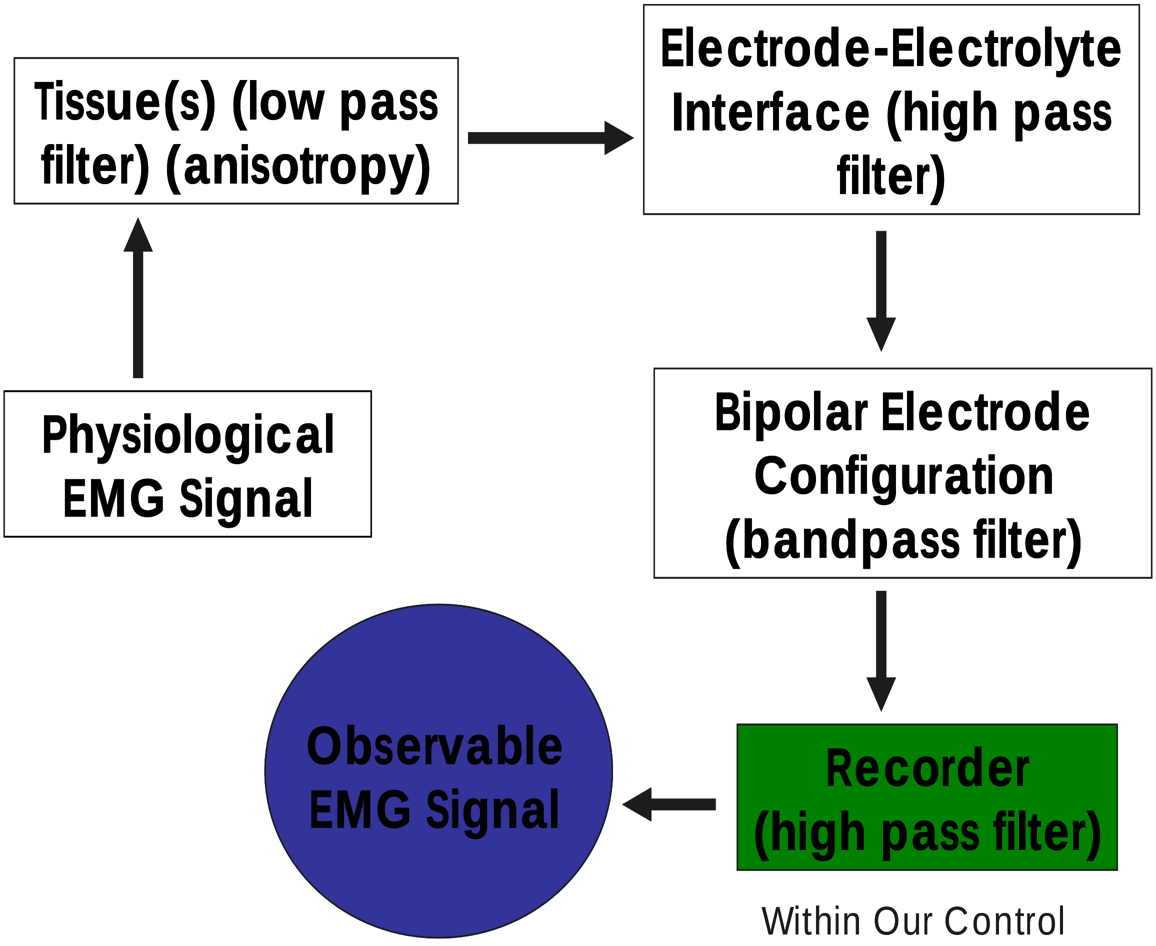


Figure 3.9. EMG Filtering For Signal Acquisition From Physiological EMG Signal To Observable EMG Signal (Recreated from *Electromyography* [De Luca, 2006]).

Electromyography (EMG) Filtering

Low Pass Filter

Filters are designed to attenuate specific frequencies or ranges of frequencies while allowing other frequencies to pass unaltered. Filters allow us to eliminate noise and other signals which might be confused with the true EMG signal. A cut off frequency must be determined (either by the EMG manufacturer or by the researchers) to determine what frequencies will be essentially ignored and which ones will be allowed to pass through. In a low pass filter, when the frequency cut off is determined, any value above the frequency cut off is attenuated to zero and is not allowed to be collected. Therefore, all EMG values will be below the frequency cut off (*Figure 3.10*).

High Pass Filter

A high pass filter works just the opposite of a low pass filter. A frequency cut off is determined and any value above it is allowed to be collected. Any value below the frequency cut off is attenuated to zero and is not collected by the EMG unit (*Figure 3.10*).

Band Pass Filter

A band pass filter allows a range of signals to be accepted. In a band pass filter, two frequency cut offs are identified (a low range and a high range) and any signal value which falls between these two frequency cut offs is accepted. Anything below the low frequency cut off and above the high frequency cut off is attenuated to zero and becomes a lost EMG signal (*Figure 3.10*).

Band Stop Filter

The band stop filter works just the opposite of a band pass filter. The band stop filter has two frequency cut offs (a low range and high range). Any signal below the low range frequency cut off or above the high range frequency cut off is allowed to be collected and any signal which falls between the two ranges is filtered out of the data by zero value attenuation. The band stop filter is most commonly used to filter out noise.

For example, a typical electrical outlet can create a 60 cycle noise effect which can be picked up in the EMG signal. In order to filter out this noise, a low frequency range of 59 Hz and a high frequency range of 61 Hz can be used to eliminate any signal from a 60 Hz electrical outlet (*Figure 3.10*).

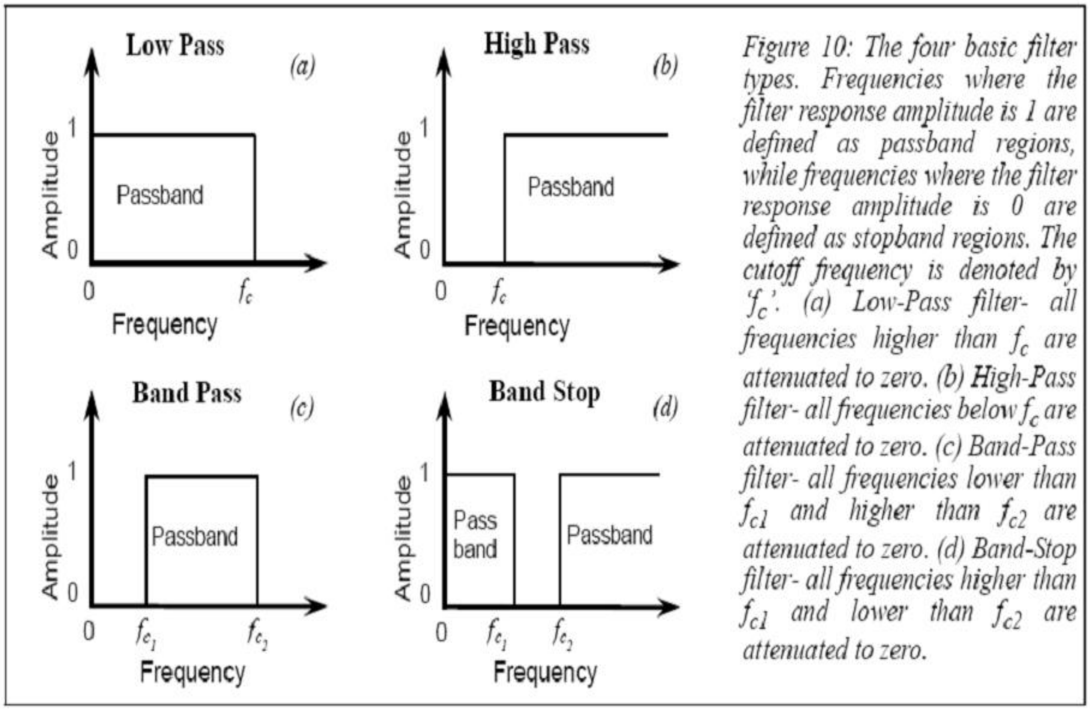


Figure 10: The four basic filter types. Frequencies where the filter response amplitude is 1 are defined as passband regions, while frequencies where the filter response amplitude is 0 are defined as stopband regions. The cutoff frequency is denoted by 'f_c'. (a) Low-Pass filter- all frequencies higher than f_c are attenuated to zero. (b) High-Pass filter- all frequencies below f_c are attenuated to zero. (c) Band-Pass filter- all frequencies lower than f_{c1} and higher than f_{c2} are attenuated to zero. (d) Band-Stop filter- all frequencies higher than f_{c1} and lower than f_{c2} are attenuated to zero.

Figure 3.10. Four Basic EMG Filters (Referenced courtesy of *Fundamental concepts in EMG signal acquisition* [De Luca, 2003]).

Electromyography (EMG) & Muscle Activation Processing

MYOLAB software was used to process raw EMG data after acquisition. EMG signals were high pass filteredwithin the FREEEMG internal hardware integrated inside each EMG probe with a cut off at 15.9 Hz (*Figure 3.11*). No other filtering was performed on the data prior to data analysis.

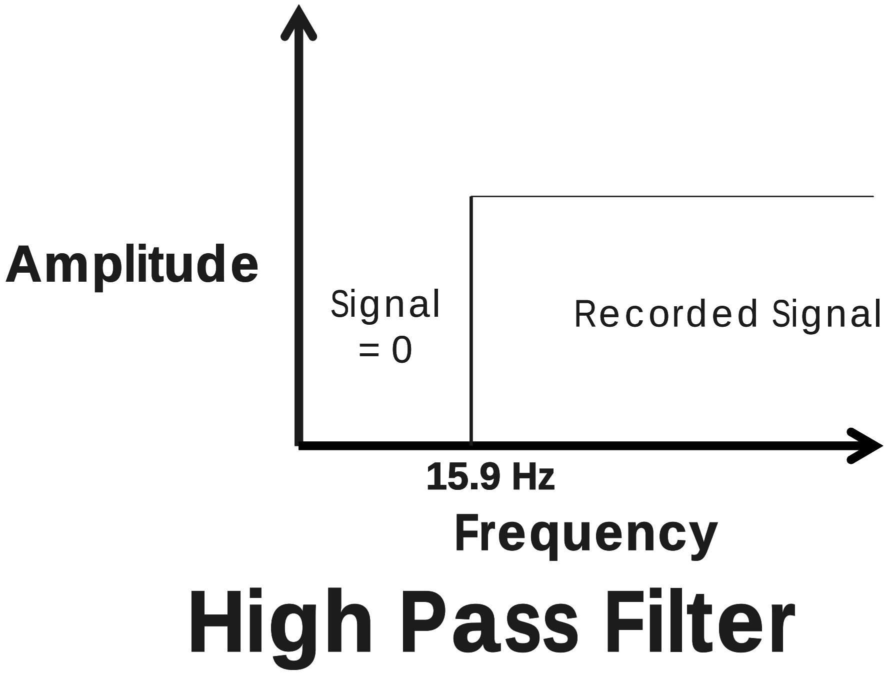


Figure 3.11. BTS FreeEMG High Pass Filter

The BTS FreeEMG system exports raw EMG signal to the computer software for analysis (*Figure 3.12).* The raw EMG was run through a root mean square (RMS) calculation with a 20 millisecond window moving average (*Figure 3.13*). The RMS EMG signal was examined to determine the average peak amplitude during jumping and cutting maneuvers (*Figure 3.14*). Squaring each data point, summing the squares, dividing the sum by the number of observations and then taking the square root can calculate the RMS. The RMS calculation is a time domain variable as the EMG signal's amplitude is measured as a function of a given time. The EMG was then normalized for comparison between participants.

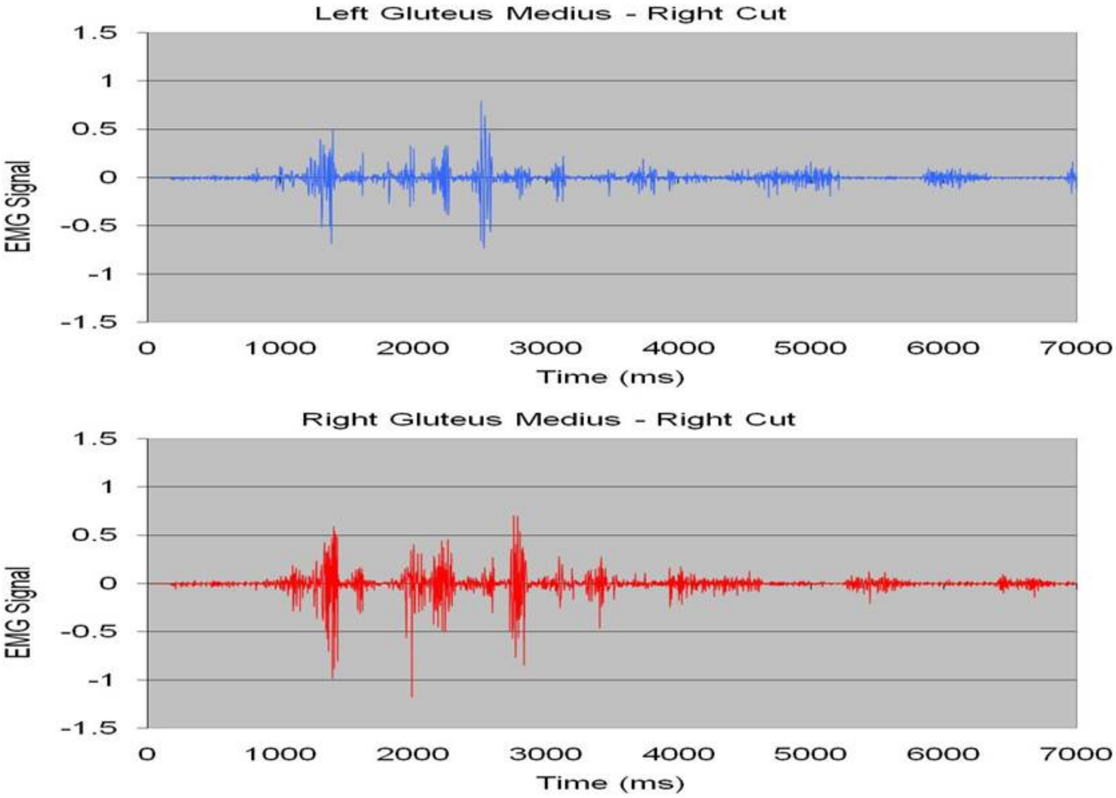


Figure 3.12. Raw EMG Data.

$$x_{\mathrm{rms}} = \sqrt{\frac{1}{T_2 - T_1}\int_{T_1}^{T_2} [f(t)]^2 \, dt}$$

Figure 3.13. Root Mean Square (RMS) Calculation.

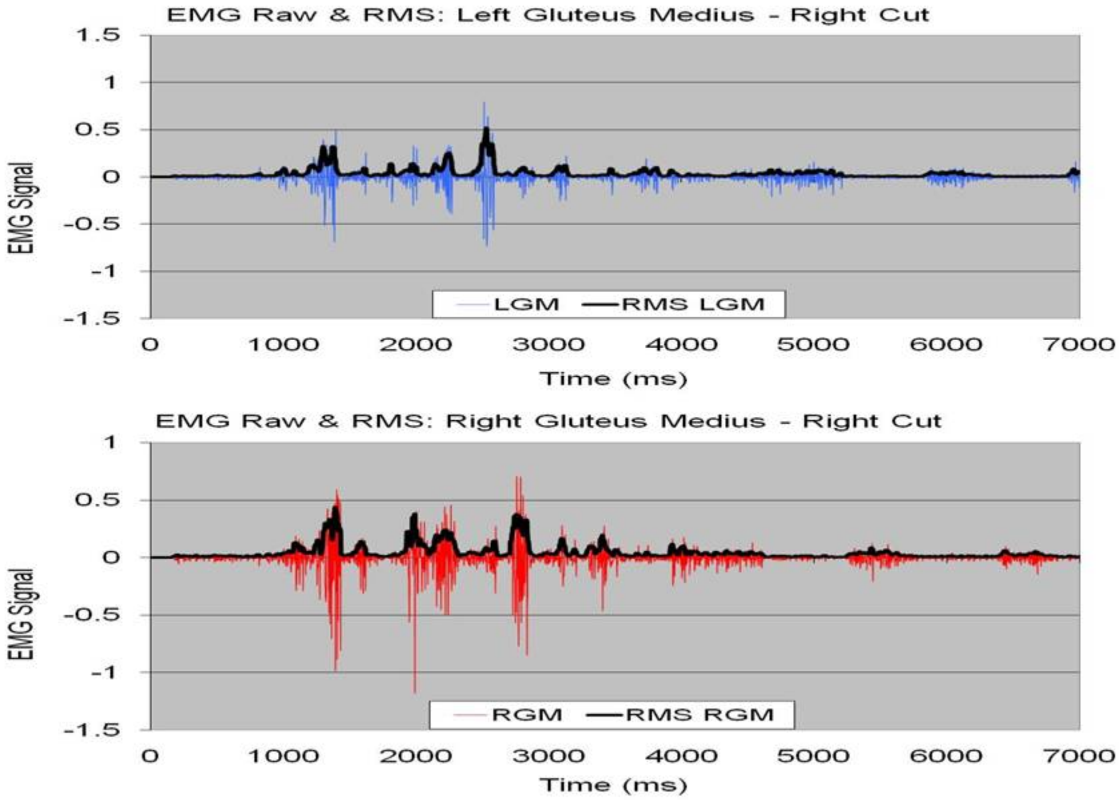


Figure 3.14. Raw EMG Data With The Root Mean Square (RMS) Calculation Overlaid.

Maximum Voluntary Isometric Contraction Normalization

Maximum voluntary isometric contraction (MVIC) is determined by calculating the peak muscle activity for each MVIC trial in the designated muscles and then taking an average across the number of trials to get the peak mean amplitude. The mean amplitudes for the MVIC trials were used to normalize the muscle activity data collected during the side-step cutting task for each respective muscle by dividing the jumping and landing EMG values by the MVIC values and then multiplying by 100. This calculation makes the normalized mean amplitude muscle activation data comparable between and among participants.

Center Run Normalization

Jumping, running and cutting are dynamic tasks and might be better represented by a dynamic normalization. The center run normalization was determined by taking the left and right cut peak mean amplitude EMG values and dividing by the peak mean amplitudes for the center run and then multiplying by 100 to make the center run normalized mean amplitude muscle activation data during the left and right land, and cut maneuver expressible as a percentage of the Center Run.

EMG Time Identification Markers

Flight Time

The flight time was calculated from the moment the subject left the pressure switch to the time the subject sustained initial contact with the forceplates (averaged over 5 trials) (*Figure 3.15*).

Push-Off Time

The time of push-off was calculated from the moment the subjects initiated contact to the time the subject left the forceplates (averaged over 5 trials) (*Figure 3.15*).

Peak Knee Flexion Time

The time to peak knee flexion angle was calculated from the initial contact, when the participant initially touches the force plate, to the time when the knee reaches peak flexion as calculated by sagittal plane kinematics (averaged over 5 trials) (*Figure 3.15*).

Peak Vertical Ground Reaction Force (vGRF) Time

The peak vertical Ground Reaction Force (vGRF) time was calculated from the initial contact to the peak in the vertical ground reaction force during the trial (averaged over 5 trials) (*Figure 3.15*).

Gluteus Medius Onset

The muscle onset timing of the gluteus medius muscle was calculated relative to the initial contact. A window of 200-300 msec prior to initial contact was used to designate muscle “quiet time” or baseline. When raw EMG values reached 3 standard deviations above the designated quite time, muscle onset were determined to occur (averaged over 5 trials) (*Figure 3.15*).

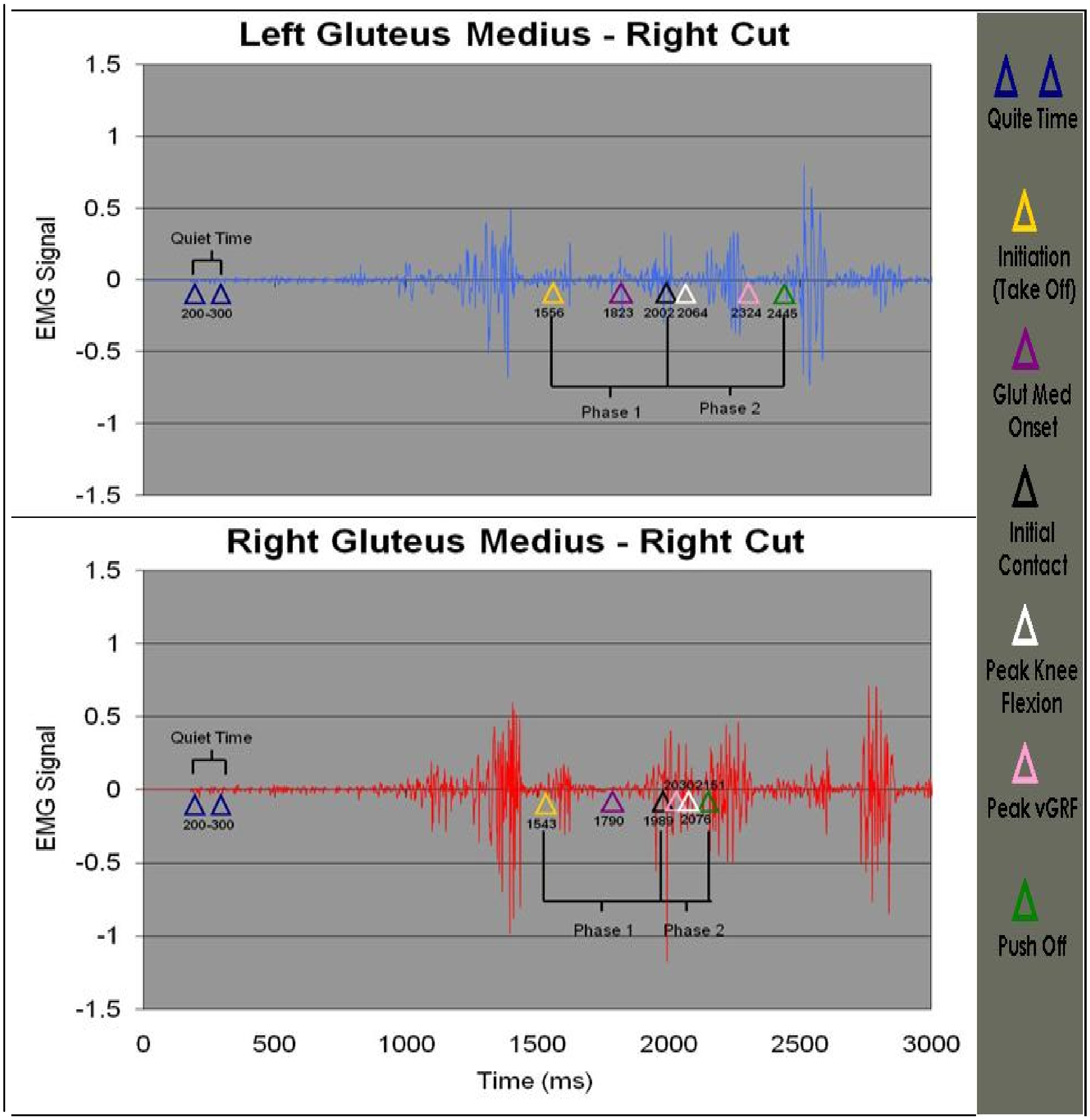


Figure 3.15. EMG Event Marker Identification.

Motion Analysis Kinematics Processing

Knee and hip kinematic data was analyzed using MATLAB. The Plug-In gate VICON marker set and model were used to analyze the kinematic data. The data was normalized from initial contact to push off as a window representing 100% of the represented time. Maximum knee flexion, hip flexion, knee valgus/varus, hip valgus/varus angles were determined for each cut direction.

Sagittal Plane Kinematics. Sagittal plane kinematics were used to determine anterior and posterior joint angles with positive angles representing joint flexion and negative angles representing joint extension. Joint angles were calculated for the hip, knee, and ankle.

Frontal Plane Kinematics. Frontal plane kinematics were used to determine abduction and adduction joint angles with positive angles representing joint abduction and negative angles representing joint adduction. Joint angles were calculated for the hip, knee, and ankle.

Transverse Plane Kinematics. Rotational kinematics were used to determine internal rotation and external rotation joint angles with positive angles representing joint internal rotation and negative angles representing joint external rotation. Joint angles were calculated for the hip, knee, and ankle.

Ground Reaction Force Processing

Peak GRF was calculated using MATLAB. Peak vertical GRF was defined as the maximum value of the VGRF (Vertical Ground Reaction Force). Body weight was used to normalize peak VGRF data. Peak sagittal (anterior/poster) and frontal (medial/lateral) was defined as the maximum value of each measure, for each the right and left leg as determined by the data collected from the force plate during ground contact.

Sagittal Plane Kinetics. Sagittal plane kinetics were used to determine anterior and posterior ground reaction forces with positive forces representing flexion and negative forces representing extension.

Frontal Plane Kinetics. Frontal plane kinetics was used to determine lateral and medial forces with positive forces representing lateral and negative forces representing medial.

Vertical Kinetics. Vertical kinematics were determined with positive values representing vertical ground reaction forces. Force values were normalized using the participant's body weight in order to compare between participants.

Benefits of the Study

There were no direct benefits to the participants in this study. However, information gained from the testing in this study will hopefully shed light on the ongoing and perplexing problem concerning the excessive numbers of lower extremity injuries, in particular ACL injuries, to the female athletic population.

Design and Analysis

The independent variable was gender. The *kinetic* dependent variables were: peak anterior/posterior ground reaction forces during landing; peak medial/lateral ground reaction forces during landing; and peak vertical ground reaction force during landing. The *kinematic* dependent variables were: joint angles at the hip and knee, including flexion, extension, abduction or adduction. Dependent measures for muscle activation included: normalized mean amplitude muscle activity (percentage of maximal voluntary isometric contraction [% MVIC]) for the right vastus medialis (RVM), left vastus medialis (LVM), right biceps femoris (RBF), left biceps femoris (LBF), right adductor longus (RAL), left adductor longus (LAL), right gluteus medius (RGM), and left gluteus medius (LGM) of both the right and left lower extremity. Dependent measures for *muscle onset times* included: time frames for the left gluteus medius and right gluteus medius.

EMG

Statistical analyses were performed using an independent samples t-test to examine the effect of gender on the EMG amplitude and time to gluteus medius muscle onset. An a-priori α level of 0.05 was set for determining statistical significance. Effect size was reported by taking the mean of the boy's scores and subtracting the mean of the girl's scores then dividing by the pooled standard deviations.

Kinematics

Statistical analyses were performed using an independent samples t-test to examine the effect of gender on the joint angles of the hip, knee, and ankle during the initial contact, peak knee flexion angles, peak ground reaction forces (GRF), and push off. An a-priori α level of 0.05 was set for determining statistical significance. Effect size was reported by taking the mean of the boy's scores and subtracting the mean of the girl's scores then dividing by the pooled standard deviations.

Kinetics

Statistical analyses were performed using an independent samples t-test to examine the effect of gender on the joint kinematics during the initial contact, peak knee flexion angles, peak ground reaction forces (GRF), and push off. An a-priori α level of 0.05 was set for determining statistical significance. Effect size was reported by taking the mean of the boy's scores and subtracting the mean of the girl's scores then dividing by the pooled standard deviations.

CHAPTER 4

Results

Electromyography (EMG)

Left Cut

MVIC Normalization

No significant differences were obtained for the EMG amplitudes (average peak means) for either the left gluteus medius (LGM) or right gluteus medius (RGM) during the first or second phase of the left cut athletic maneuver (*Table 4.1*).

Center Run Dynamic Normalization

No significant differences were obtained for the EMG amplitudes (average peak means) for either the LGM or RGM during the first or second phase of the left cut athletic maneuver (*Table 4.1*).

Gluteus Medius Muscle Onset

None of the EMG muscle onsets were statistically different between the male and female basketball athletes (*Table 4.2*).

Center Cut

During the center cut, participants were free to choose the lead leg for the cut. The female participants were a little more consistent than the males were when choosing a lead leg. The females were more likely to choose to lead with the left leg (66.67% of the time) compared to the males who were more likely to lead with the right leg (56.25% of the time) (*Figure 4.1*).

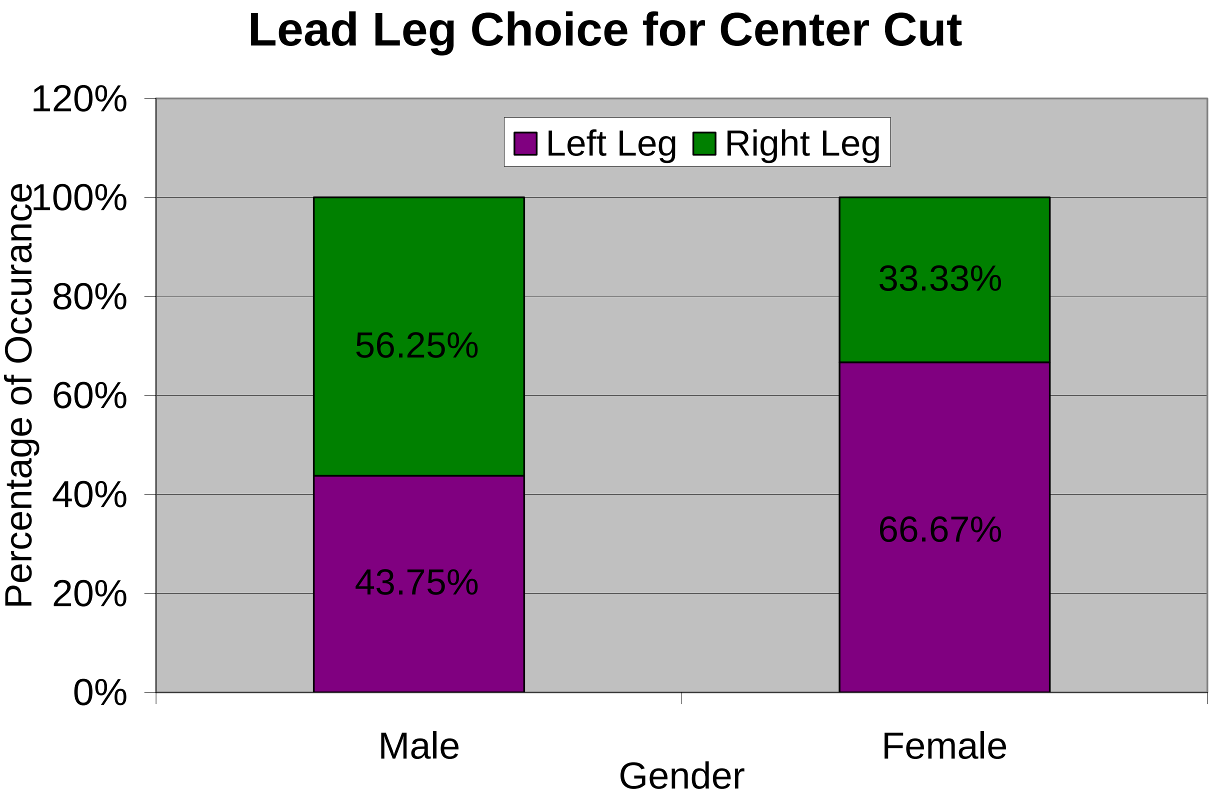


Figure 4.1. Choice In The Lead Leg For The Center Cut Based on Gender Across All The Trials Analyzed.

During the center cut, there was a great deal of variance between and among participants when self-selecting a lead leg for the center cut. The participants were given no formal instructions on the center cut, only on the left and right cuts (which were performed utilizing a side-step cutting maneuver). Six of the ten total participants (three male and three female) self-selected a lead leg and remained consistent throughout the trials. The other four participants self-selected random lead leg patterns throughout their trials (*Figure 4.2* & *Figure 4.3*).

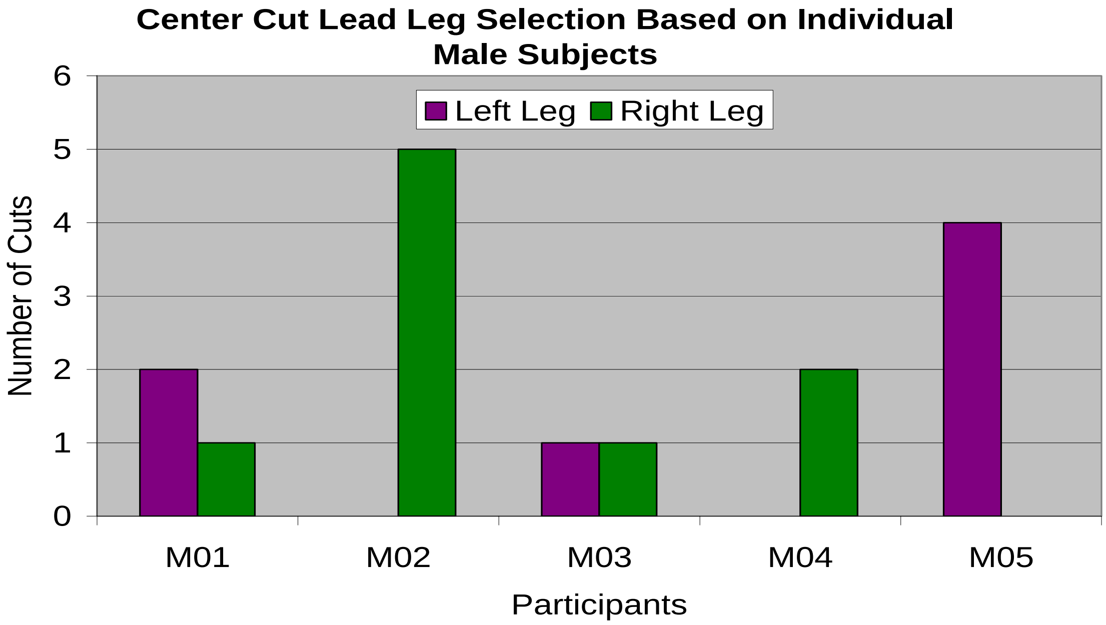


Figure 4.2. Male Participants Center Cut Lead Leg Self-Selection

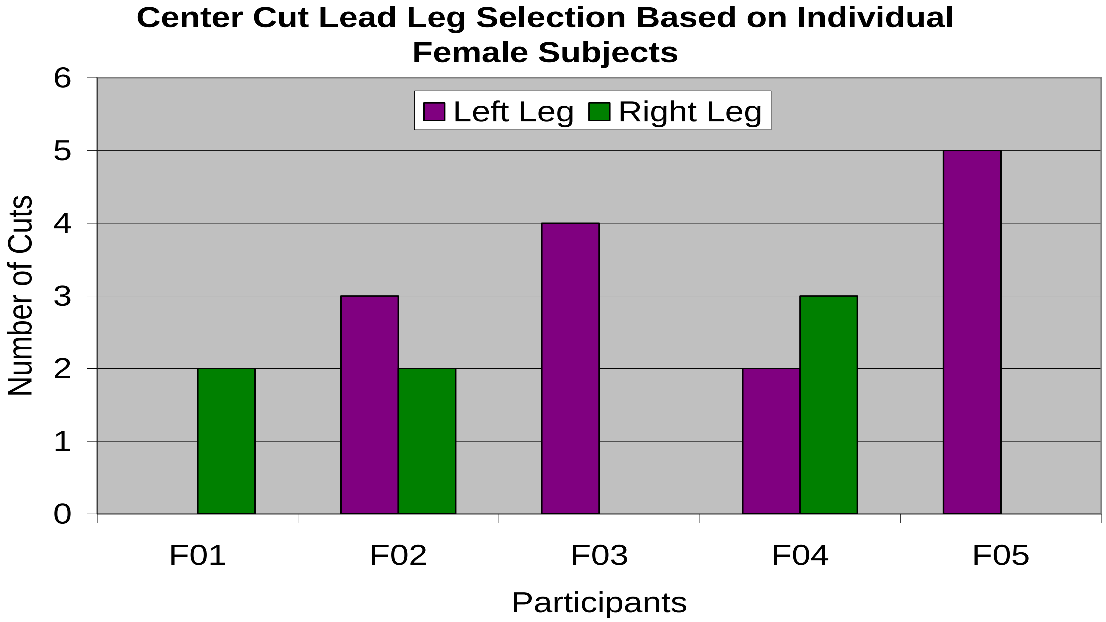


Figure 4.3. Female Participants Center Cut Lead Leg Self-Selection

MVIC Normalization

No significant differences were obtained for the EMG amplitudes (average peak means) for either the LGM or RGM during the first or second phase of the straight run athletic maneuver (*Table 4.1*).

Gluteus Medius Muscle Onset

None of the EMG muscle onsets were statistically different between the male and female basketball athletes (*Table 4.2*).

Right Cut

MVIC Normalization

No significant differences were obtained for the EMG amplitudes (average peak means) for either the LGM or RGM during the first or second phase of the right cut athletic maneuver (*Table 4.1*).

Center Run Dynamic Normalization

No significant differences were obtained for the EMG amplitudes (average peak means) for either the LGM or RGM during the first or second phase of the right cut athletic maneuver (*Table 4.1*).

Gluteus Medius Muscle Onset

None of the EMG muscle onsets were statistically different between the male and female basketball athletes (*Table 4.2*).

EMG Normalization

This study is one of a hand full which has used a dynamic measure as a means of normalizing EMG values. Cowling and Steele (2001) were one of the first to suggest a dynamic motion might be a better normalization method for comparing dynamic movements because we are essentially comparing the same types of movements instead of comparing a static (MVIC normalization) to a dynamic movement. Although there were no statistically significant results between the genders, the novelty of the analysis should be examined. In many of the graphs (*Figure 4.4, Figure 4.6, Figure 4.7, Figure*

4.9) comparing the Maximum Voluntary Isometric Contraction (MVIC) normalization and the center run normalization, the center run normalization means and standard deviations. A decrease in the range of standard deviation means that the tasks are becoming more specific and are more reproducible on the part of the participant. It is not uncommon for values of the MVIC normalization during dynamic contractions to be 200% or 300% of the MVIC value. Theoretically, values in the center cut run should be closer to 100% because the movement is dynamic and not a static movement trying to evaluate a dynamic movement.

In *Figure 4.4* the MVIC normalization is compared with the center run normalization for the left gluteus medius during phase I. Phase I was classified from the initiation of the jump to the initial contact with the force plates. This graph has some very large values within the percentage (%) values in the center run normalization. The mean for the center run normalization in the left cut is 381% ± 620; however, upon closer inspection, one of the female participants had a 1489% normalized EMG value during the center run. It appears there is an issue in the EMG data collection during this participant's trial.

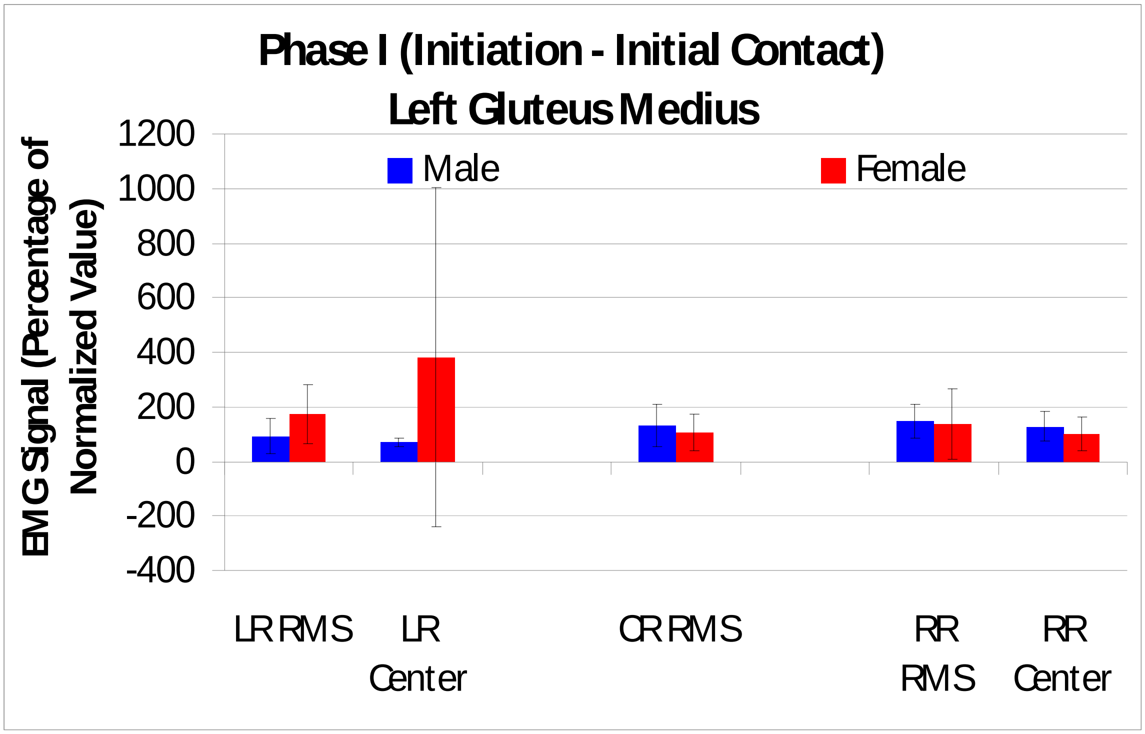


Figure 4.4. Phase I Normalized EMG Values For The Left Gluteus Medius Based On The MVICs and On The Center Run With The RMS Indicating A MVIC Normalization And The Center Indicating A Center Run (Straight Cut) Normalization.

A test was run to determine if the value seen in the center run normalization for the females left gluteus medius during the left cut reached a value to be excluded as an outlier. A Grubbs' test (extreme studentized deviate) test was performed to determine outlier status with $p = 0.05$. The Grubbs' test calculates the robustness of the data and determines if data falls within three or fewer standard deviations away from the mean. An outlier was detected for the 1489% value ($p < 0.05$, $Z = 1.788$). When the value was removed from the data set, the mean and standard deviation were removed and now follow the trends of the right gluteus medius with the center cut normalized values means and standard deviations being smaller and more compact compared with the MVIC normalization. For the female's center run normalization the mean drops from 380.97 to 103.74 and the standard deviation drops from 620.36 to 27.40 with the removal of the outlier. *Figure 4.5* represents the original data as expressed in *Figure 4.4* and also represents the means and standard deviations with the one outlier removed.

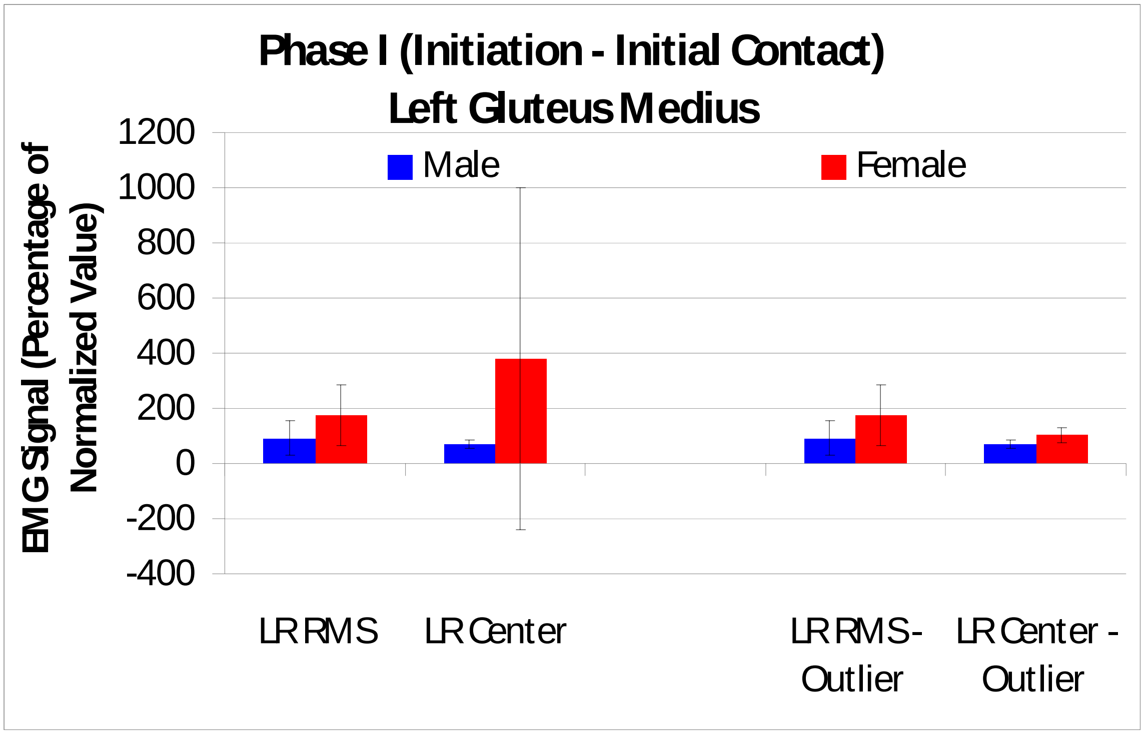


Figure 4.5. Phase I Normalized EMG Values For The Left Gluteus Medius Based On The MVICs and On The Center Run With The RMS Indicating A MVIC Normalization And The Center Indicating A Center Run (Straight Cut) Normalization.

In *Figure 4.6*, the normalized values expressed a trend similar to what was expected. The LR RMS, CR RMS and The RR RMS are the MVIC normalizations and the LR Center and RR Center are the Center Cut Normalizations. The normalized values are for the right gluteus medius during phase I of the trial. When comparing the center run normalization, we can see a decrease in the means across both the left and right cutting direction in both genders, and we see the standard deviations drop and become a much smaller range.

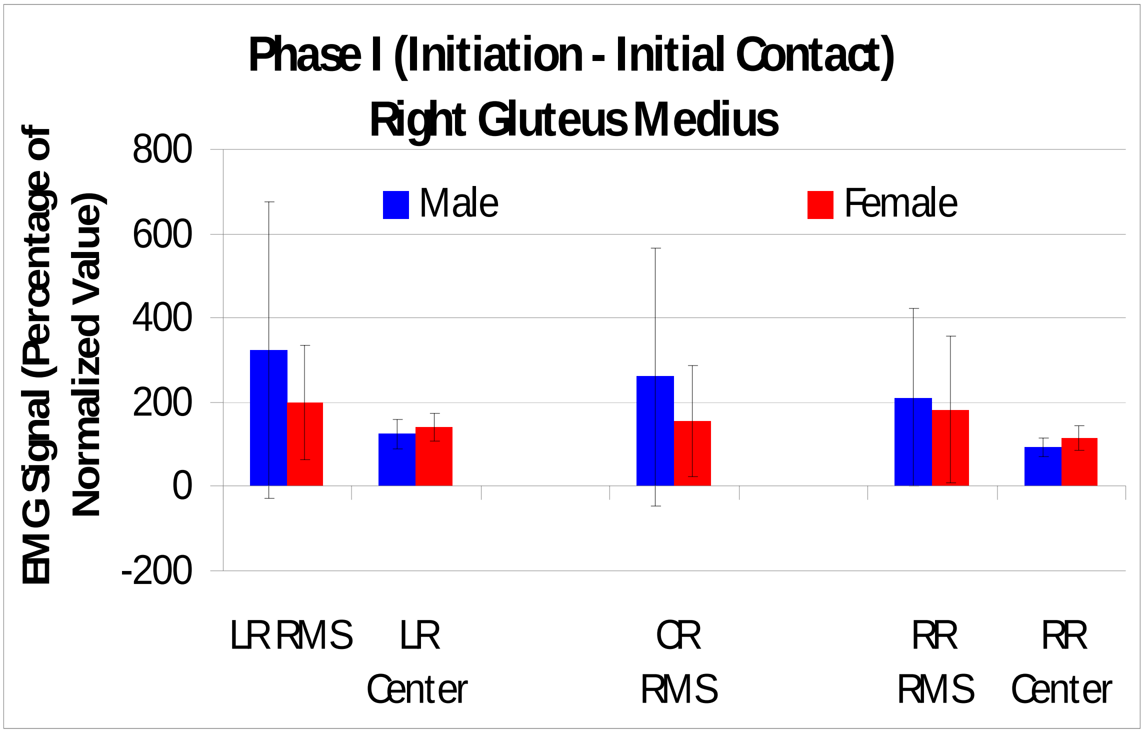


Figure 4.6. Phase I Normalized EMG Values For The Right Gluteus Medius Based On The MVICs and On The Center Run With The RMS Indicating A MVIC Normalization And The Center Indicating A Center Run (Straight Cut) Normalization.

In *Figure 4.7*, EMG values are compared again using both the MVIC and Center Cut Normalizations this time during Phase II. Phase II was classified from initial contact with the force plate to push off of the cut. This graph too has some very large values in the center run normalization in the left cut.

The mean for the center run normalization in the left cut is 1555% ± 3263; however, upon closer inspection, one of the female participants had a 7393% normalized EMG value during the center run. It appears there is an issue in the EMG data collection during this participant's trial.

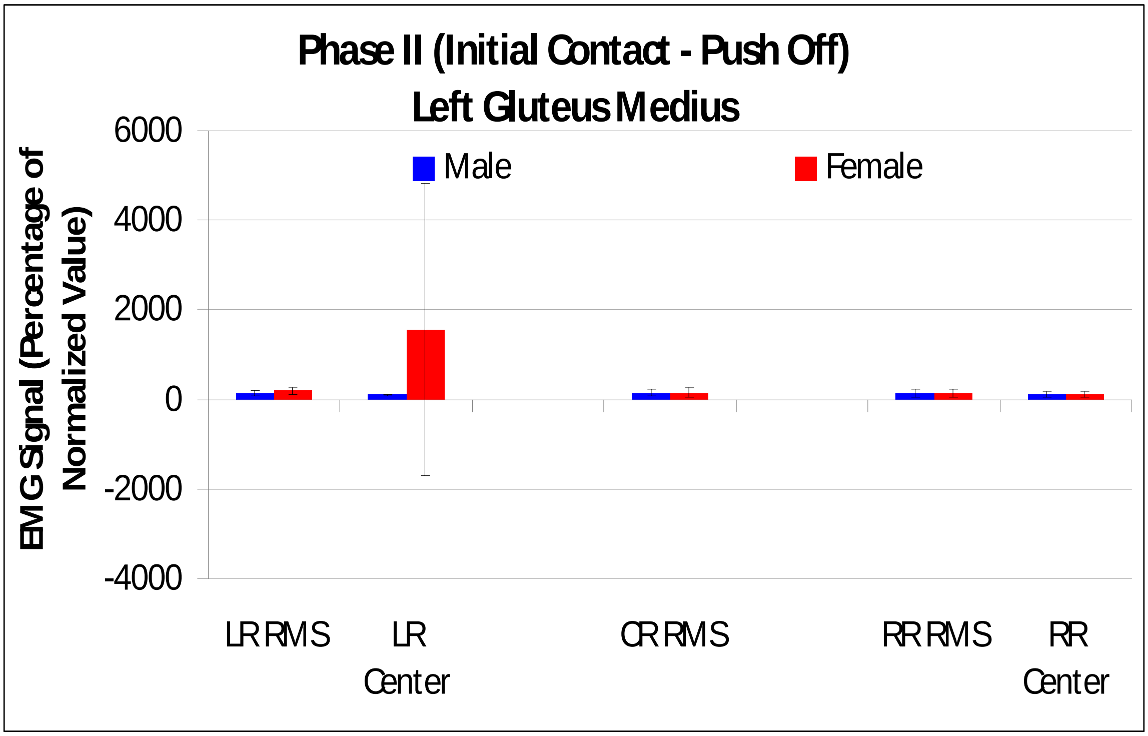


Figure 4.7. Phase II Normalized EMG Values For The Left Gluteus Medius Based On The MVICs and On The Center Run With The RMS Indicating A MVIC Normalization And The Center Indicating A Center Run (Straight Cut) Normalization.

A test was run to determine if the value seen in the center run normalization for the females left gluteus medius during the left cut reached a value to be excluded as an outlier. A Grubbs' test (extreme studentized deviate) test was performed to determine outlier status with $p = 0.05$. The Grubbs' test calculates the robustness of the data and determines if data falls within three or fewer standard deviations away from the mean. An outlier was detected for the 7392.73% value ($p < 0.05$, $Z = 1.788$). When the value was removed from the data set, the mean and standard deviation were removed and now follow the trends of the right gluteus medius with the center cut normalized values means and standard deviations being smaller and more compact compared with the MVIC normalization. For the female's center run normalization the mean drops from 1555.24 to 95.87 and the standard deviation drops from 3263.27 to 11.57 with the removal of the outlier. *Figure 4.8* represents the original data as expressed in *Figure 4.7* and also represents the means and standard deviations with the one outlier removed.

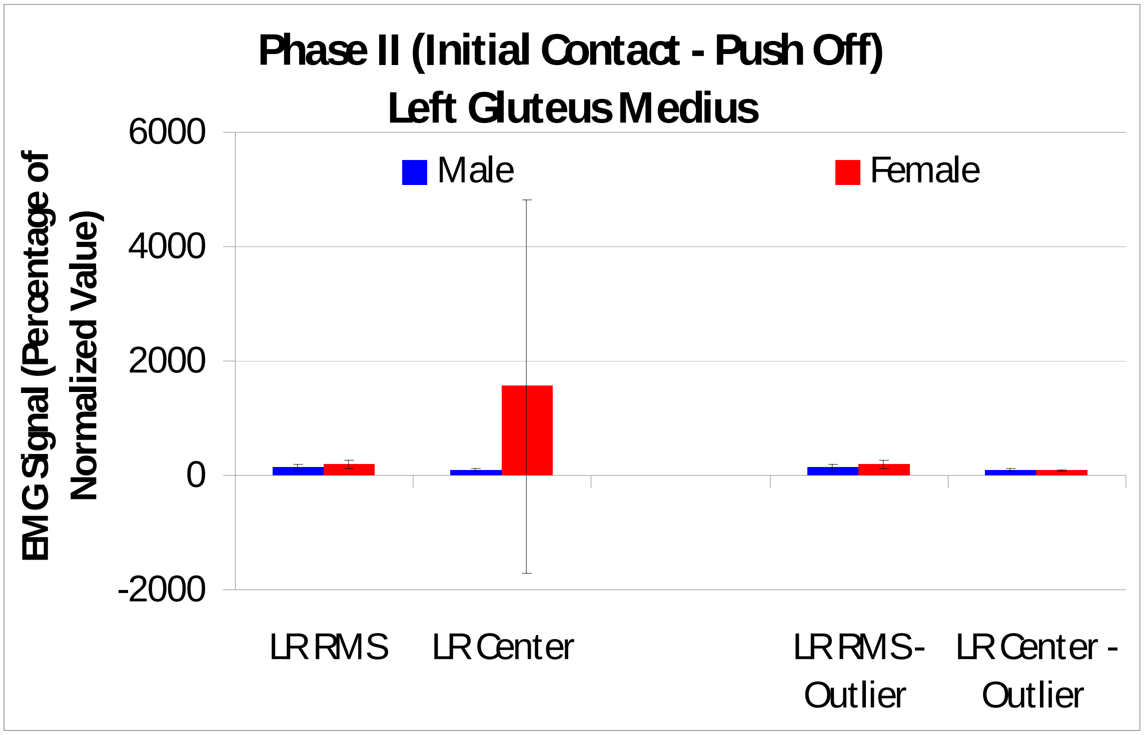


Figure 4.8. Phase I Normalized EMG Values For The Left Gluteus Medius Based On The MVICs and On The Center Run With The RMS Indicating A MVIC Normalization And The Center Indicating A Center Run (Straight Cut) Normalization.

In *Figure 4.9*, the normalized values expressed a trend similar to what was expected. The LR RMS, CR RMS and The RR RMS are the MVIC normalizations and the LR Center and RR Center are the Center Cut Normalizations. The normalized values are for the right gluteus medius during phase II of the trial. When comparing the center run normalization, we can see a decrease in the means across both the left and right cutting direction in both genders, and we see the standard deviations drop and become a much smaller range.

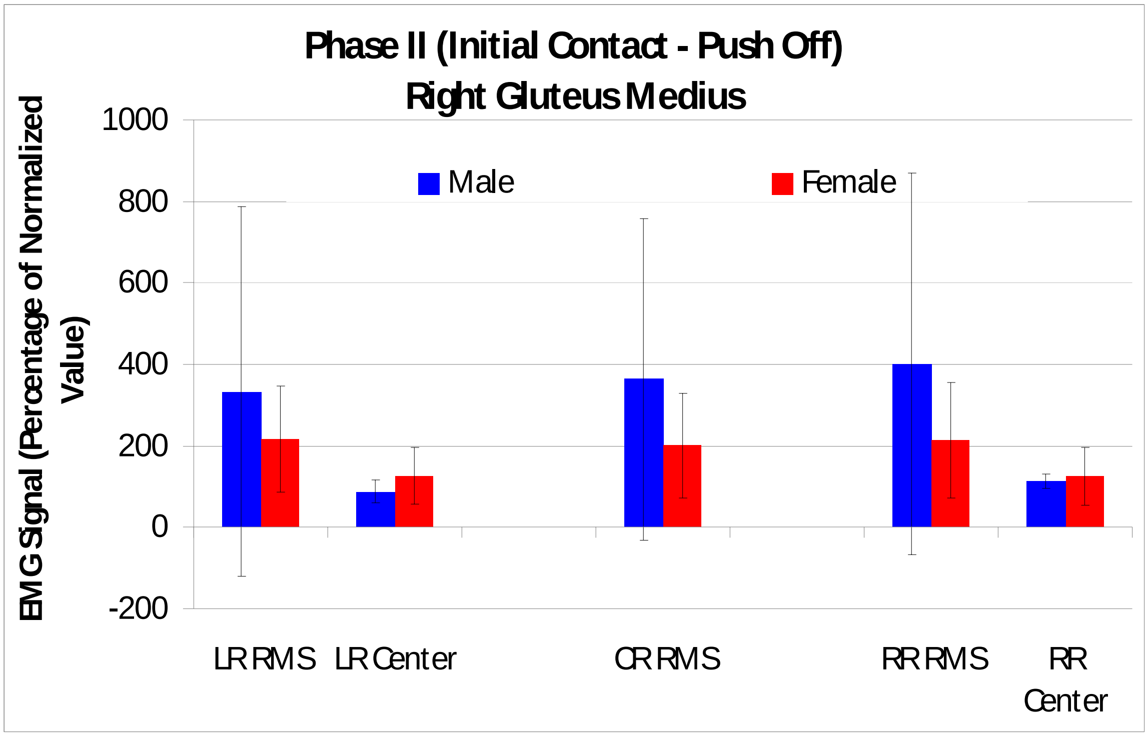


Figure 4.9. Phase II Normalized EMG Values For The Right Gluteus Medius Based On The MVICs and On The Center Run With The RMS Indicating A MVIC Normalization And The Center Indicating A Center Run (Straight Cut) Normalization.

Table 4.1

Electromyography (EMG) Values Calculated During Testing Presented As Percentages of Normalized EMG Signal.

Left Cut	Left Gluteus Medius						Right Gluteus Medius					
	Male		*Female*				*Male*		*Female*			
Phase 1	% Mean	St. Dev	%Mean	St. Dev	T-test	SE	% Mean	St. Dev	%Mean	St. Dev	T-test	SE
Left Run Maximum Average - Based on RMS	92.31	64.08	174.95	108.46	0.18	-0.48	324.13	350.31	199.88	134.45	0.48	0.26
Left Run Maximum Average - Based on Center Run	69.26	15.73	380.97	620.36	0.29	-0.49	125.63	34.63	140.35	32.90	0.51	-0.22
Phase 2	% Mean	St. Dev	%Mean	St. Dev	T-test	SE	% Mean	St. Dev	%Mean	St. Dev	T-test	SE
Left Run Maximum Average - Based on RMS	135.26	61.38	191.95	76.90	0.23	-0.41	333.26	453.68	215.56	129.90	0.59	0.20
Left Run Maximum Average - Based on Center Run	96.85	21.73	1555.24	3263.27	0.35	-0.44	88.13	27.42	126.25	68.37	0.28	-0.40

Center Cut	Left Gluteus Medius						Right Gluteus Medius					
	Male		*Female*				*Male*		*Female*			
Phase 1	% Mean	St. Dev	%Mean	St. Dev	T-test	SE	% Mean	St. Dev	%Mean	St. Dev	T-test	SE
Center Run Maximum Average - Based on RMS	130.58	76.92	105.22	65.82	0.59	0.18	261.15	305.59	156.70	131.63	0.50	0.24
Phase 2	% Mean	St. Dev	%Mean	St. Dev	T-test	SE	% Mean	St. Dev	%Mean	St. Dev	T-test	SE
Center Run Maximum Average - Based on RMS	142.45	75.84	149.50	106.49	0.91	-0.04	363.22	394.67	200.53	128.52	0.41	0.31

Right Cut	Left Gluteus Medius						Right Gluteus Medius					
	Male		*Female*				*Male*		*Female*			
Phase 1	% Mean	St. Dev	%Mean	St. Dev	T-test	SE	% Mean	St. Dev	%Mean	St. Dev	T-test	SE
Right Run Maximum Average - Based on RMS	145.11	62.02	135.90	130.28	0.89	0.05	211.07	210.53	182.78	175.40	0.82	0.07
Right Run Maximum Average - Based on Center Run	128.03	55.49	101.14	61.84	0.49	0.23	92.66	20.69	114.33	28.57	0.21	-0.44
Phase 2	% Mean	St. Dev	%Mean	St. Dev	T-test	SE	% Mean	St. Dev	%Mean	St. Dev	T-test	SE
Right Run Maximum Average - Based on RMS	145.51	83.31	150.91	91.70	0.92	-0.03	401.07	468.04	214.18	141.70	0.42	0.31
Right Run Maximum Average - Based on Center Run	108.01	56.49	108.88	66.69	0.98	-0.01	112.64	17.95	124.23	70.57	0.73	-0.13

🟨 = Stastical Significance ($p < 0.05$)

Table 4.2

Electromyography (EMG) Onset Values Calculated During Testing Presented In Milliseconds.

Left Cut

Left Gluteus Medius						**Right Gluteus Medius**					
Male		Female				Male		Female			
Mean (ms)	St. Dev	Mean (ms)	St. Dev	Ttest	SE	Mean (ms)	St. Dev	Mean (ms)	St. Dev	Ttest	SE
-167.94	66.79	-184.17	13.90	0.61	0.20	-167.99	64.01	-184.79	21.72	0.61	0.20

Center Cut

Left Gluteus Medius						**Right Gluteus Medius**					
Male		Female				Male		Female			
Mean (ms)	St. Dev	Mean (ms)	St. Dev	Ttest	SE	Mean (ms)	St. Dev	Mean (ms)	St. Dev	Ttest	SE
-156.74	79.60	-185.68	13.86	0.45	0.31	-156.78	78.47	-190.68	7.12	0.45	0.40

Right Cut

Left Gluteus Medius						**Right Gluteus Medius**					
Male		Female				Male		Female			
Mean (ms)	St. Dev	Mean (ms)	St. Dev	Ttest	SE	Mean (ms)	St. Dev	Mean (ms)	St. Dev	Ttest	SE
-183.41	30.23	-174.02	27.91	0.62	-0.16	-167.64	49.98	-185.54	15.70	0.62	0.27

🟨 = Stastical Significance ($p < 0.05$)

Kinematics

Left Cut

Hip

No statistical differences were noted between the genders for the kinematics of the left or right hip during initial contact (*Table 4.3*), peak knee flexion (*Table 4.6*), peak vertical ground reaction forces (vGRF) (*Table 4.9*), or push off (*Table 4.12*).

Knee

No statistical differences were noted between the genders for the kinematics of the left or right knee during initial contact (*Table 4.4*), peak knee flexion (*Table 4.7*), peak vGRF (*Table 4.10*), or push off (*Table 4.13*).

Ankle

Statistical differences ($p = 0.019$, $d = -1.06$) were noted in the between the genders for the kinematics of the left ankle during peak knee flexion angles with the female athletes sustaining greater dorsiflexion (anterior) angles (22.29° ± 1.71°) than their male counterparts (14.13° ± 5.95°) (*Figure 4.10, Table 4.8*). In addition the male participants had much larger standard deviations, which could suggest variance between male subjects. No differences were noted between the genders for the right ankle during peak knee flexion angles (*Table 4.8*). No statistical differences were noted between the genders for the kinematics of the left or right ankle during initial contact (*Table 4.5*), peak vGRF (*Table 4.11*), or push off (*Table 4.14*).

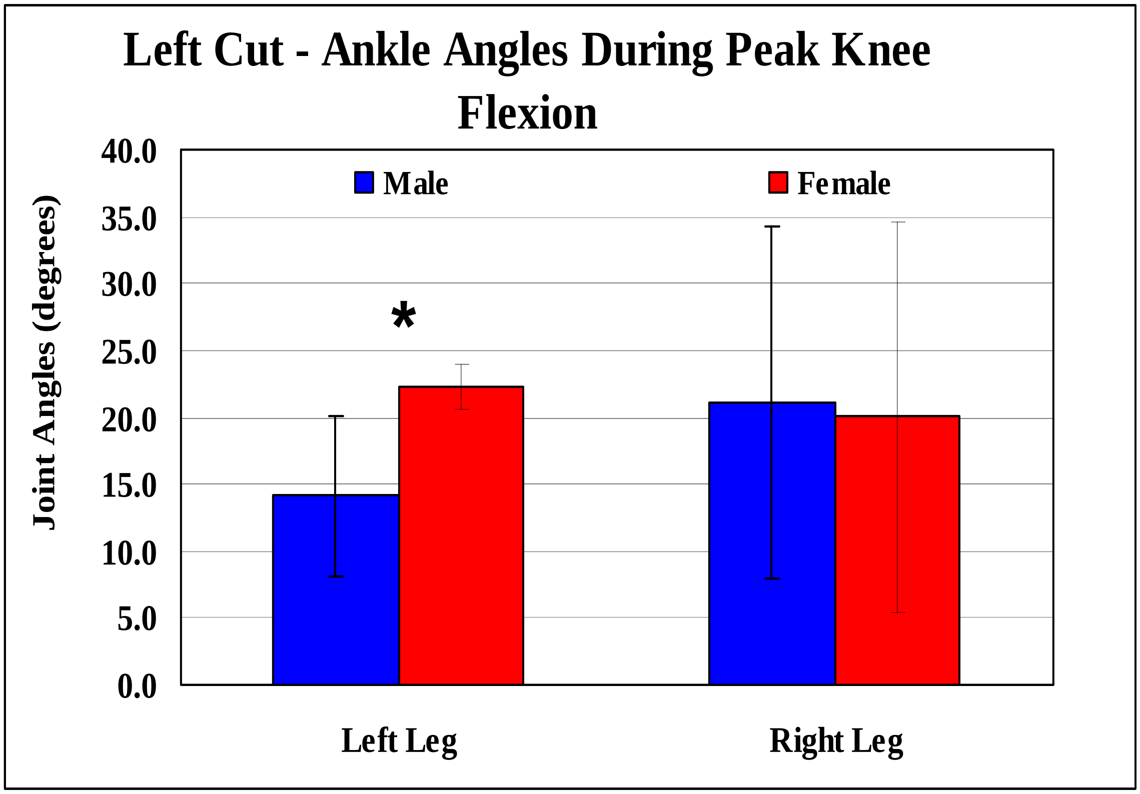


Figure 4.10. Statistically Significant Differences In The Left Leg While Cutting Left With Positive Values Indicating Dorsiflexion And Negative Values Indicating Plantarflexion.

Center Cut

Hip

No statistical differences were noted between the genders for the kinematics of the left or right hip during initial contact (*Table 4.3*), peak knee flexion (*Table 4.6*), peak vGRF (*Table 4.9*), or push off (*Table 4.12*).

Knee

No statistical differences were noted between the genders for the kinematics of the left or right knee during initial contact (*Table 4.4*), peak knee flexion (*Table 4.7*), peak vGRF (*Table 4.10*), or push off (*Table 4.13*).

Ankle

Statistical differences (p = 0.012, *d* = 1.03) were noted in the between the genders for the kinematics of the right ankle during peak GRF with the male athletes sustaining greater dorsiflexion (anterior) angles (22.61° ± 2.62°) than their female counterparts (17.75° ± 2.09°) (*Figure 4.11, Table 4.11*). No differences were noted between the genders for the left ankle during peak knee flexion angles (*Table 4.11*). No statistical differences were noted between the genders for the kinematics of the left or right ankle during initial contact (*Table 4.5*), peak knee flexion (*Table 4.8*), or push off (*Table 4.14*).

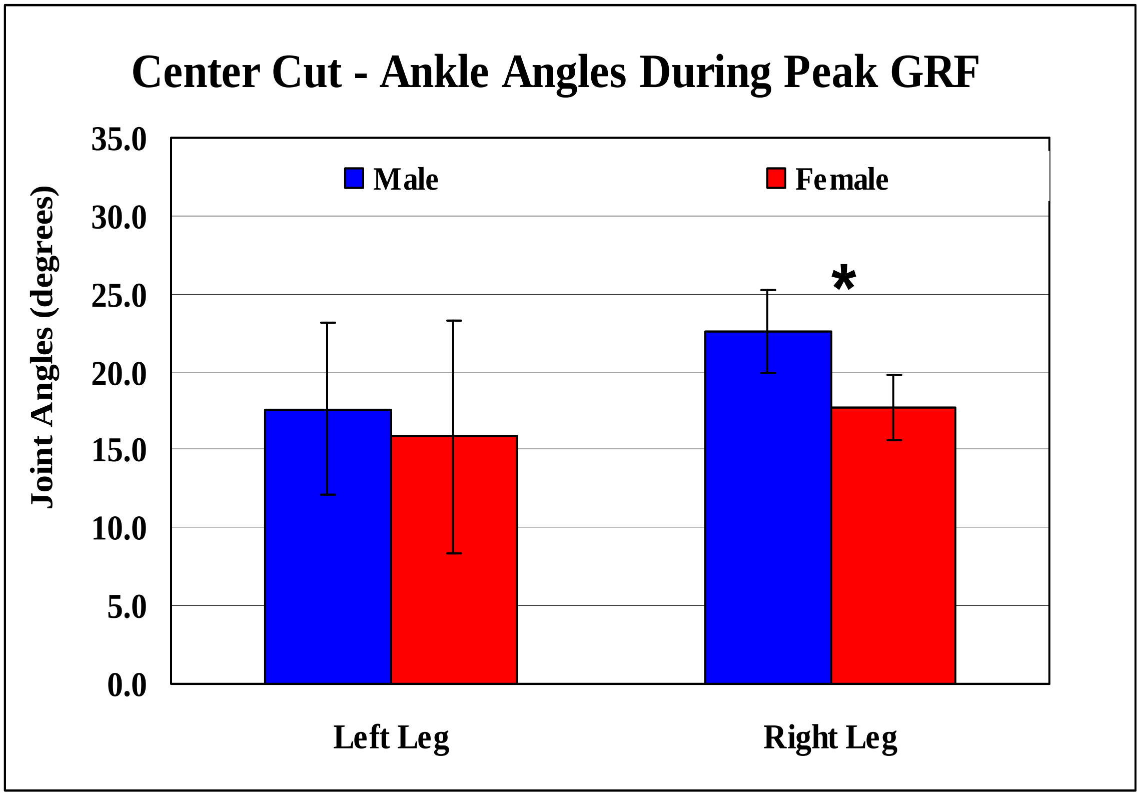


Figure 4.11. Statistically Significant Differences In The Right Leg While Performing A Straight Run With Positive Values Indicating Dorsiflexion And Negative Values Indicating Plantarflexion.

Right Cut

Hip

No statistical differences were noted between the genders for the kinematics of the left or right hip during initial contact (*Table 4.2*), peak knee flexion (*Table 4.6*), peak vGRF (*Table 4.9*), or push off (*Table 4.12*).

Knee

No statistical differences were noted between the genders for the kinematics of the left or right knee during initial contact (*Table 4.4*), peak knee flexion (*Table 4.7*), peak vGRF (*Table 4.10*), or push off (*Table 4.13*).

Ankle

No statistical differences were noted between the genders for the kinematics of the left or right ankle during initial contact (*Table 4.5*), peak knee flexion (*Table 4.8*), peak vGRF (*Table 4.11*), or push off (*Table 4.14*).

Table 4.3

Kinematic Values Of The Hip Joint Calculated During Initial Contact.

Initial Contact - Hip Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	27.15	24.88	-5.93	-11.30	-0.10	-4.47
		SD	8.68	4.91	2.92	2.49	15.45	11.40
	Female	Mean	28.32	26.30	-6.50	-7.65	-10.99	-5.18
		SD	5.22	4.40	5.74	5.63	17.85	6.58
		T-test	0.80	0.64	0.85	0.22	0.33	0.91
		SE	-0.08	-0.15	0.07	-0.45	0.33	0.04
Center Cut	Male	Mean	27.15	24.88	-5.93	-11.30	-0.10	-4.47
		SD	8.68	4.91	2.92	2.49	15.45	11.40
	Female	Mean	28.32	26.30	-6.50	-7.65	-10.99	-5.18
		SD	5.22	4.40	5.74	5.63	17.85	6.58
		T-test	0.80	0.64	0.85	0.22	0.33	0.91
		SE	-0.08	-0.15	0.07	-0.45	0.33	0.04
Right Cut	Male	Mean	25.79	26.94	-11.11	-5.91	-0.22	-6.84
		SD	3.77	2.16	4.27	1.99	15.65	13.07
	Female	Mean	29.03	29.60	-7.27	-7.91	-12.29	-5.46
		SD	4.03	4.72	4.88	6.01	16.97	6.71
		T-test	0.22	0.29	0.22	0.50	0.28	0.84
		SE	-0.42	-0.39	-0.42	0.25	0.37	-0.07

🟨 = Stastical Significance ($p < 0.05$)

Table 4.4

Kinematic Values Of The Knee Joint Calculated During Initial Contact.

Initial Contact - Knee Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	21.38	19.37	4.86	3.42	-2.11	-2.80
		SD	7.01	2.16	10.54	5.38	9.40	5.53
	Female	Mean	23.46	18.77	-2.58	-1.19	7.91	-0.67
		SD	7.36	2.29	6.77	5.19	8.43	6.83
		T-test	0.66	0.69	0.22	0.21	0.11	0.60
		SE	-0.14	0.13	0.43	0.44	-0.56	-0.17
Center Cut	Male	Mean	21.38	19.37	4.86	3.42	-2.11	-2.80
		SD	7.01	2.16	10.54	5.38	9.40	5.53
	Female	Mean	23.46	18.77	-2.58	-1.19	7.91	-0.67
		SD	7.36	2.29	6.77	5.19	8.43	6.83
		T-test	0.66	0.69	0.22	0.21	0.11	0.60
		SE	-0.14	0.13	0.43	0.44	-0.56	-0.17
Right Cut	Male	Mean	17.43	23.49	4.48	1.89	-4.25	0.49
		SD	6.96	4.85	8.58	5.72	11.31	10.10
	Female	Mean	21.76	23.50	-3.03	-0.30	5.80	-0.27
		SD	4.57	4.72	6.30	4.93	4.11	7.58
		T-test	0.28	1.00	0.15	0.53	0.10	0.90
		SE	-0.38	0.00	0.50	0.21	-0.65	0.04

🟨 = Stastical Significance ($p < 0.05$)

Table 4.5

Kinematic Values Of The Ankle Joint Calculated During Initial Contact.

Initial Contact - Ankle Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	-1.25	-4.29	0.69	-0.07	-1.78	-0.16
		SD	8.04	5.08	2.66	2.77	10.12	10.08
	Female	Mean	-4.59	-6.89	-1.56	-1.51	7.52	7.05
		SD	7.07	4.67	3.71	2.21	11.97	9.19
		T-test	0.51	0.42	0.30	0.39	0.22	0.27
		SE	0.22	0.27	0.35	0.29	-0.42	-0.37
Center Cut	Male	Mean	-1.25	-4.29	0.69	-0.07	-1.78	-0.16
		SD	8.04	5.08	2.66	2.77	10.12	10.08
	Female	Mean	-4.59	-6.89	-1.56	-1.51	7.52	7.05
		SD	7.07	4.67	3.71	2.21	11.97	9.19
		T-test	0.51	0.42	0.30	0.39	0.22	0.27
		SE	0.22	0.27	0.35	0.29	-0.42	-0.37
Right Cut	Male	Mean	-5.20	-1.95	1.34	-0.69	-4.37	1.52
		SD	7.86	7.42	2.04	3.84	7.38	12.32
	Female	Mean	-7.22	-3.29	-1.95	-0.85	8.90	4.72
		SD	4.56	5.28	3.61	1.63	11.79	7.02
		T-test	0.63	0.75	0.11	0.93	0.07	0.63
		SE	0.16	0.11	0.58	0.03	-0.69	-0.17

🟨 = Stastical Significance ($p < 0.05$)

Table 4.6

Kinematic Values Of The Hip Joint Calculated During Peak Knee Flexion.

Peak Knee Flexion - Hip Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	38.16	23.34	-10.39	-12.13	1.54	-6.05
		SD	14.75	13.17	5.11	8.02	20.27	7.52
	Female	Mean	30.48	17.03	-7.37	-4.45	-8.39	-5.85
		SD	8.30	19.26	4.26	6.45	22.26	9.38
		T-test	0.34	0.56	0.34	0.13	0.48	0.97
		SE	0.33	0.19	-0.32	-0.53	0.23	-0.01
Center Cut	Male	Mean	23.50	17.19	-6.18	-7.86	-0.02	-9.32
		SD	10.69	7.84	2.21	2.44	18.51	10.11
	Female	Mean	27.36	19.71	-5.31	-6.67	1.68	-7.46
		SD	9.28	18.87	7.63	7.66	9.60	8.81
		T-test	0.56	0.79	0.81	0.75	0.86	0.76
		SE	-0.19	-0.09	-0.09	-0.12	-0.06	-0.10
Right Cut	Male	Mean	30.39	27.28	-13.12	-5.78	2.19	-5.82
		SD	7.12	4.98	8.91	3.74	19.31	13.22
	Female	Mean	27.40	31.29	-4.70	-8.26	-8.87	-7.30
		SD	7.01	9.08	6.40	7.27	20.57	11.84
		T-test	0.52	0.41	0.12	0.52	0.41	0.86
		SE	0.21	-0.28	-0.55	0.23	0.28	0.06

🟨 = Stastical Significance ($p < 0.05$)

Table 4.7

Kinematic Values Of The Knee Joint Calculated During Peak Knee Flexion.

Peak Knee Flexion - Knee Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	50.79	42.09	6.12	-3.64	3.60	4.01
		SD	15.34	13.04	21.18	4.50	10.48	9.71
	Female	Mean	53.03	37.74	-7.08	-6.21	8.27	1.24
		SD	15.13	18.88	14.69	6.76	5.42	10.01
		T-test	0.82	0.68	0.29	0.50	0.40	0.67
		SE	-0.07	0.14	0.37	0.23	-0.29	0.14
Center Cut	Male	Mean	45.97	42.06	2.16	-4.75	7.43	3.38
		SD	8.79	13.99	15.96	3.76	11.54	9.88
	Female	Mean	48.41	38.97	-8.73	-6.14	9.07	1.48
		SD	12.44	20.77	12.79	7.70	4.87	10.24
		T-test	0.73	0.79	0.27	0.73	0.78	0.77
		SE	-0.11	0.09	0.38	0.12	-0.10	0.09
Right Cut	Male	Mean	45.71	40.37	2.74	0.61	7.30	3.90
		SD	4.51	6.17	15.60	7.88	8.86	12.08
	Female	Mean	44.96	49.52	-7.07	-6.21	9.41	2.42
		SD	7.44	9.01	13.61	10.65	2.17	8.99
		T-test	0.85	0.10	0.32	0.28	0.62	0.83
		SE	0.06	-0.60	0.34	0.37	-0.19	0.07

🟨 = Stastical Significance ($p < 0.05$)

Table 4.8

Kinematic Values Of The Ankle Joint Calculated During Peak Knee Flexion.

Peak Knee Flexion - Ankle Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	14.13	21.05	2.99	0.11	-11.04	-0.11
		SD	5.95	13.16	2.31	2.53	7.94	9.75
	Female	Mean	22.29	20.06	1.53	-0.59	-4.16	2.72
		SD	1.71	14.61	3.70	3.42	12.77	11.94
		T-test	0.02	0.91	0.48	0.72	0.34	0.69
		SE	-1.06	0.04	0.24	0.12	-0.33	-0.13
Center Cut	Male	Mean	15.54	17.22	1.89	0.59	-6.28	-2.78
		SD	11.10	11.47	1.99	2.98	6.32	10.23
	Female	Mean	21.19	15.06	0.58	-0.24	-0.89	1.86
		SD	9.38	16.07	3.55	2.61	13.11	9.80
		T-test	0.41	0.81	0.49	0.65	0.43	0.48
		SE	-0.28	0.08	0.24	0.15	-0.28	-0.23
Right Cut	Male	Mean	25.32	20.38	1.60	1.88	-6.07	-7.72
		SD	5.69	4.02	1.52	1.90	7.07	8.39
	Female	Mean	24.72	25.70	0.66	0.96	-1.41	-2.40
		SD	4.47	4.42	2.98	1.23	11.59	4.17
		T-test	0.86	0.08	0.54	0.39	0.46	0.24
		SE	0.06	-0.63	0.21	0.29	-0.25	-0.42

🟨 = Stastical Significance ($p < 0.05$)

Table 4.9

Kinematic Values Of The Hip Joint Calculated During Peak GRF.

Peak Ground Reaction Forces (GRF) - Hip Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	27.18	25.21	-6.04	-11.27	-1.37	-8.44
		SD	11.88	8.73	4.41	5.68	17.36	13.84
	Female	Mean	27.14	24.53	-6.59	-4.99	-11.77	-4.27
		SD	6.17	7.36	5.82	6.59	19.25	8.42
		T-test	1.00	0.90	0.87	0.15	0.40	0.58
		SE	0.00	0.04	0.05	-0.51	0.28	-0.19
Center Cut	Male	Mean	17.96	21.03	-5.05	-6.45	-0.34	-9.07
		SD	8.14	7.20	4.13	2.20	19.19	11.26
	Female	Mean	28.06	27.99	-6.78	-6.63	-2.20	-5.87
		SD	6.50	4.50	6.03	8.63	9.45	6.06
		T-test	0.06	0.10	0.61	0.96	0.85	0.59
		SE	-0.69	-0.59	0.17	0.02	0.06	-0.18
Right Cut	Male	Mean	25.09	24.48	-11.87	-5.21	-0.12	-7.86
		SD	3.77	2.26	5.79	3.18	17.89	13.09
	Female	Mean	26.63	28.75	-4.62	-8.03	-9.48	-9.64
		SD	7.24	6.50	6.46	7.05	20.74	8.74
		T-test	0.68	0.20	0.10	0.44	0.47	0.81
		SE	-0.14	-0.49	-0.59	0.28	0.24	0.08

🟨 = Stastical Significance ($p < 0.05$)

Table 4.10

Kinematic Values Of The Knee Joint Calculated During Peak GRF.

Peak Ground Reaction Forces (GRF) - Knee Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (º)	Flex-Ext (º)	AB-AD (º)	AB-AD (º)	IR-ER (º)	IR-ER (º)
Left Cut	Male	Mean	36.36	40.01	3.56	-3.47	3.26	5.36
		SD	11.02	6.28	15.81	7.71	9.58	7.26
	Female	Mean	36.01	39.92	-7.71	-4.97	10.19	1.00
		SD	6.05	4.86	12.17	8.35	6.02	7.95
		T-test	0.95	0.98	0.24	0.78	0.21	0.39
		SE	0.02	0.01	0.40	0.09	-0.44	0.29
Center Cut	Male	Mean	32.78	41.73	2.58	-4.15	4.96	4.58
		SD	9.32	12.21	15.36	4.21	13.61	9.60
	Female	Mean	34.17	38.41	-8.53	-5.29	8.88	0.78
		SD	9.49	3.84	11.48	8.07	3.43	8.26
		T-test	0.82	0.58	0.23	0.79	0.55	0.52
		SE	-0.07	0.21	0.41	0.09	-0.23	0.21
Right Cut	Male	Mean	35.71	33.53	2.23	0.55	5.02	3.82
		SD	6.51	4.83	16.11	6.37	11.40	11.32
	Female	Mean	41.53	38.58	-7.35	-6.64	9.73	1.96
		SD	8.70	4.07	12.41	9.51	3.67	8.12
		T-test	0.27	0.11	0.32	0.20	0.40	0.77
		SE	-0.38	-0.57	0.34	0.45	-0.31	0.10

🟨 = Stastical Significance ($p < 0.05$)

Table 4.11

Kinematic Values Of The Ankle Joint Calculated During Peak GRF.

Peak Ground Reaction Forces (GRF) - Ankle Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	15.42	23.64	2.10	0.54	-7.41	-3.23
		SD	3.10	5.72	3.24	3.06	12.07	11.13
	Female	Mean	17.77	20.78	-0.03	0.58	1.68	-1.01
		SD	6.43	4.85	2.46	1.46	8.12	6.29
		T-test	0.48	0.42	0.28	0.98	0.20	0.71
		SE	-0.25	0.27	0.37	-0.01	-0.45	-0.13
Center Cut	Male	Mean	17.62	22.61	1.94	1.10	-7.05	-5.16
		SD	5.49	2.62	1.71	2.83	6.28	10.17
	Female	Mean	15.88	17.75	-0.72	0.06	4.51	0.68
		SD	7.44	2.09	3.03	1.25	9.56	6.53
		T-test	0.68	0.01	0.13	0.47	0.05	0.31
		SE	0.13	1.03	0.56	0.25	-0.73	-0.35
Right Cut	Male	Mean	21.27	16.28	2.41	0.98	-8.64	-4.84
		SD	6.66	4.58	1.75	3.10	5.22	11.76
	Female	Mean	23.39	18.14	0.56	-0.07	-1.04	1.31
		SD	4.81	6.21	3.31	0.80	12.77	4.65
		T-test	0.58	0.60	0.30	0.48	0.25	0.31
		SE	-0.18	-0.17	0.36	0.27	-0.42	-0.37

🟨 = Stastical Significance ($p < 0.05$)

Table 4.12

Kinematic Values Of The Hip Joint Calculated During Push Off.

Push Off - Hip Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	26.37	20.90	-8.39	-12.73	-0.02	-3.57
		SD	18.91	11.03	4.20	7.34	20.65	9.21
	Female	Mean	31.10	13.60	-7.01	-4.88	-8.25	-5.69
		SD	8.87	18.74	4.64	6.31	22.31	9.51
		T-test	0.63	0.47	0.64	0.11	0.56	0.73
		SE	-0.17	0.25	-0.16	-0.57	0.19	0.11
Center Cut	Male	Mean	14.17	20.14	-7.70	-8.05	-1.03	-10.05
		SD	10.16	13.06	3.50	2.96	18.83	12.96
	Female	Mean	18.78	19.25	-6.14	-6.99	3.21	-4.85
		SD	17.92	18.52	6.80	7.59	9.06	11.96
		T-test	0.63	0.93	0.66	0.78	0.66	0.53
		SE	-0.16	0.03	-0.15	-0.10	-0.15	-0.21
Right Cut	Male	Mean	16.85	35.41	-14.00	-7.20	-0.04	-4.98
		SD	15.75	12.28	9.45	3.81	18.91	12.00
	Female	Mean	13.77	33.40	-7.23	-7.74	-8.29	-6.12
		SD	19.59	6.78	4.85	5.92	20.78	11.91
		T-test	0.79	0.76	0.19	0.87	0.53	0.88
		SE	0.09	0.11	-0.47	0.06	0.21 #	0.05

🟨 = Stastical Significance ($p < 0.05$)

Table 4.13

Kinematic Values Of The Knee Joint Calculated During Push Off.

Push Off - Knee Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	39.72	40.41	5.38	-1.49	2.46	3.11
		SD	20.99	12.32	21.39	6.79	11.43	10.57
	Female	Mean	52.79	33.83	-6.30	-5.99	8.39	0.76
		SD	15.14	18.78	15.17	6.85	5.19	9.98
		T-test	0.29	0.53	0.35	0.33	0.32	0.73
		SE	-0.36	0.21	0.32	0.33	-0.36	0.11
Center Cut	Male	Mean	38.14	47.20	0.72	-5.46	6.28	4.40
		SD	11.67	18.54	16.23	7.48	12.55	10.60
	Female	Mean	42.12	41.49	-8.10	-3.60	6.65	0.36
		SD	19.61	21.76	13.08	9.03	8.64	9.95
		T-test	0.71	0.67	0.37	0.73	0.96	0.55
		SE	-0.13	0.14	0.30	-0.11	-0.02	0.20
Right Cut	Male	Mean	33.63	56.00	0.42	1.47	3.18	3.78
		SD	13.65	16.07	14.99	7.13	14.39	11.72
	Female	Mean	29.53	59.25	-5.37	-5.61	3.96	-0.50
		SD	17.90	11.12	14.05	11.37	7.23	8.65
		T-test	0.69	0.72	0.55	0.27	0.92	0.53
		SE	0.13	-0.12	0.20	0.38	-0.04	0.21

🟨 = Stastical Significance ($p < 0.05$)

Table 4.14

Kinematic Values Of The Ankle Joint Calculated During Push Off.

Push Off - Ankle Angles

			Left	Right	Left	Right	Left	Right
			Flex-Ext (°)	Flex-Ext (°)	AB-AD (°)	AB-AD (°)	IR-ER (°)	IR-ER (°)
Left Cut	Male	Mean	5.83	17.51	1.71	1.02	-6.36	-3.11
		SD	20.84	14.13	4.75	3.47	17.63	12.02
	Female	Mean	20.85	15.61	1.57	-0.58	-4.33	2.71
		SD	4.70	14.65	3.69	3.42	12.72	11.95
		T-test	0.15	0.84	0.96	0.48	0.84	0.46
		SE	-0.59	0.07	0.02	0.23	-0.07	-0.24
Center Cut	Male	Mean	7.43	14.60	0.79	1.25	-2.21	-4.86
		SD	19.61	12.44	3.28	3.25	12.13	10.84
	Female	Mean	13.32	12.45	0.33	-0.16	0.33	1.57
		SD	20.51	15.79	3.59	2.68	13.43	10.12
		T-test	0.65	0.82	0.84	0.47	0.76	0.36
		SE	-0.15	0.08	0.07	0.24	-0.10	-0.31
Right Cut	Male	Mean	10.09	15.25	0.50	2.11	-1.45	-7.68
		SD	24.28	9.59	2.59	2.13	10.21	7.88
	Female	Mean	8.72	17.08	-0.58	1.02	3.36	-2.54
		SD	23.58	9.98	2.49	1.79	10.59	6.16
		T-test	0.93	0.78	0.52	0.41	0.49	0.28
		SE	0.03	-0.09	0.21	0.28	-0.23	-0.37

🟨 = Stastical Significance ($p < 0.05$)

Kinetics

Left Cut

Statistical differences (p = 0.022, *d* = 0.93) were noted between the genders for the left foot ground reaction forces with the male participants incurring anterior forces (0.64 N/kg ± 0.58 N/kg) and female participants sustaining posterior forces (-0.89 N/kg ± 1.06 N/kg) during peak knee flexion angles (*Figure 4.12*, *Table 4.16*). No statistical differences were noted between the genders for the right foot ground reaction forces during peak knee flexion angles (*Table 4.16*).

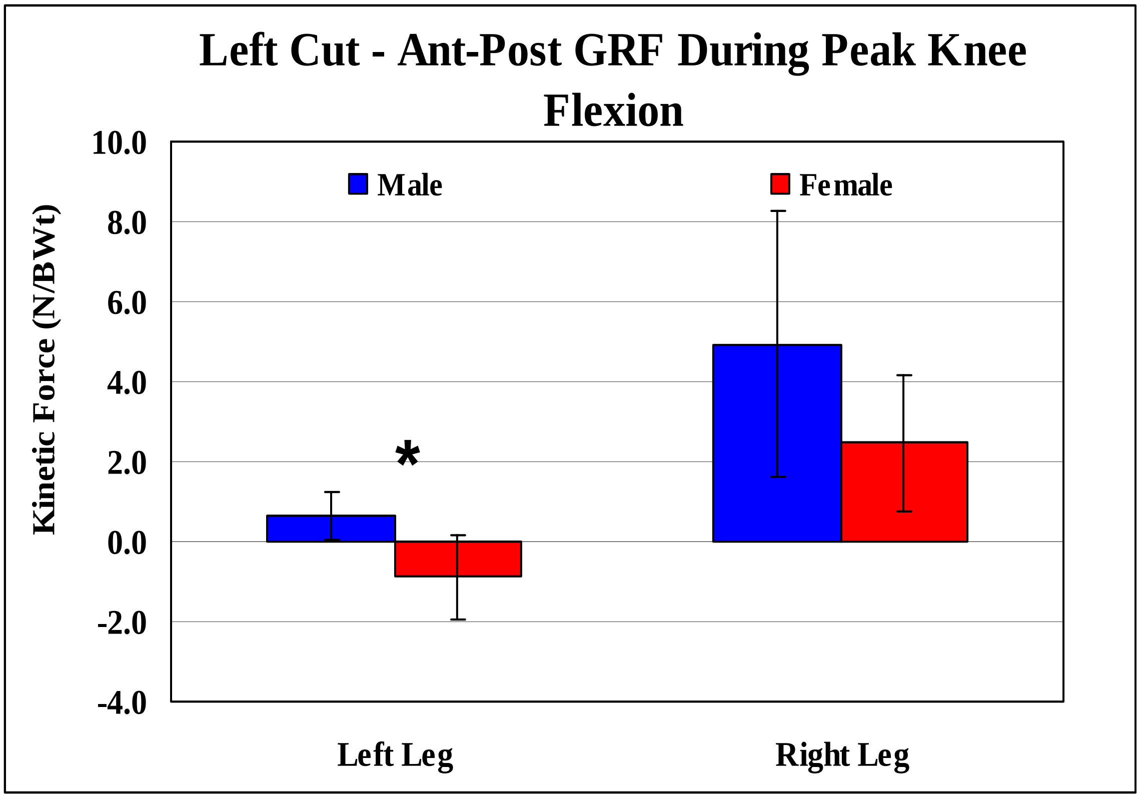


Figure 4.12. Statistically Significant Differences Anterior-Posterior Ground Reaction Forces In The Left Leg While Performing A Left Cut With Positive Values Indicating Anterior And Negative Values Indicating Posterior.

Statistical differences (p = 0.040, *d* = 0.81) were noted between the genders for the left foot ground reaction forces with the male participants incurring anterior forces (0.53 N/kg ± 0.58 N/kg) and female participants sustaining posterior forces (-0.84 N/kg ± 1.11 N/kg) during the push off (*Figure 4.13*, *Table 4.18*). No statistical differences were noted between the genders for the right foot ground reaction forces during the push off (*Table 4.18*).

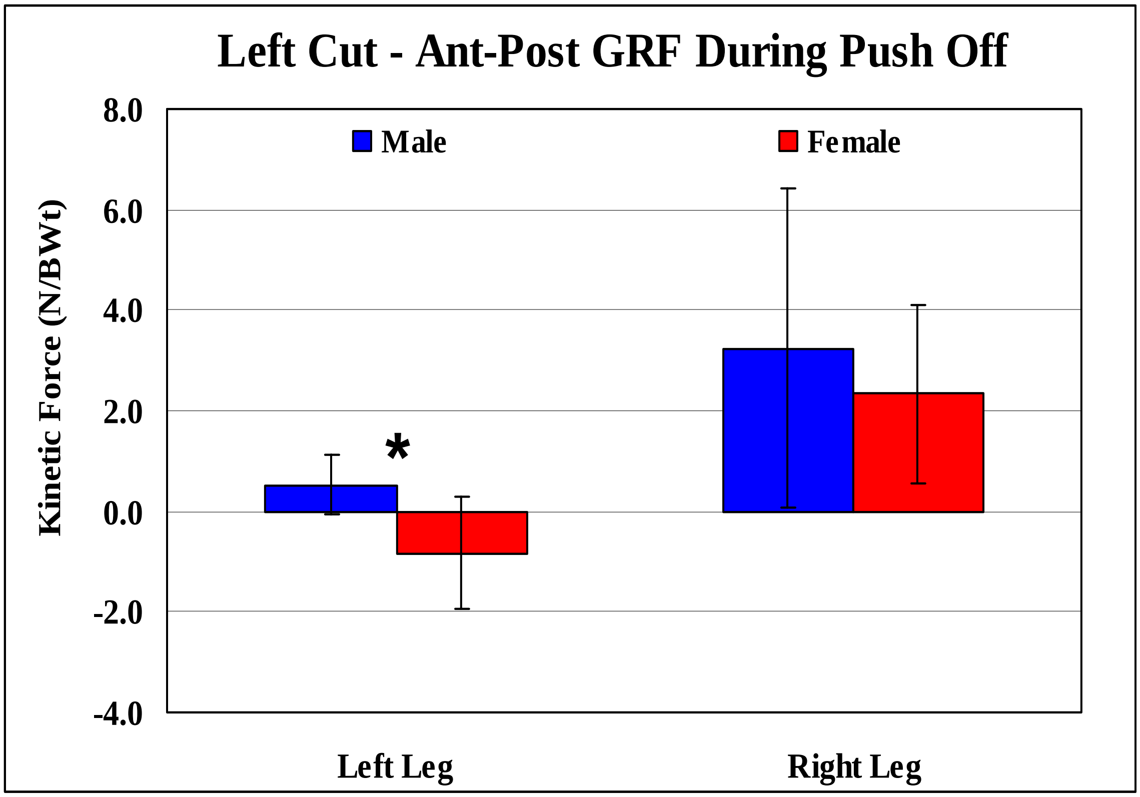


Figure 4.13. Statistically Significant Differences Flexion-Extension Ground Reaction Forces In The Left Leg While Performing A Left Cut With Positive Values Indicating Anterior And Negative Values Indicating Posterior.

No statistical differences were noted between the genders for the kinetics of the left or right foot during initial contact (*Table 4.15*) or peak GRF (*Table 4.17*).

Center Cut

Statistical differences (p = 0.010, *d* = -1.06) were noted between the genders for the right foot ground reaction forces with the male participants incurring lateral forces (-2.189N ± 1.188N) and female participants sustaining medial forces (0.620N ± 1.451N) during peak GRF (*Figure 4.14, Table 4.17*). No statistical differences were noted between the genders for the left foot ground reaction forces during peak GRF (*Table 4.17*).

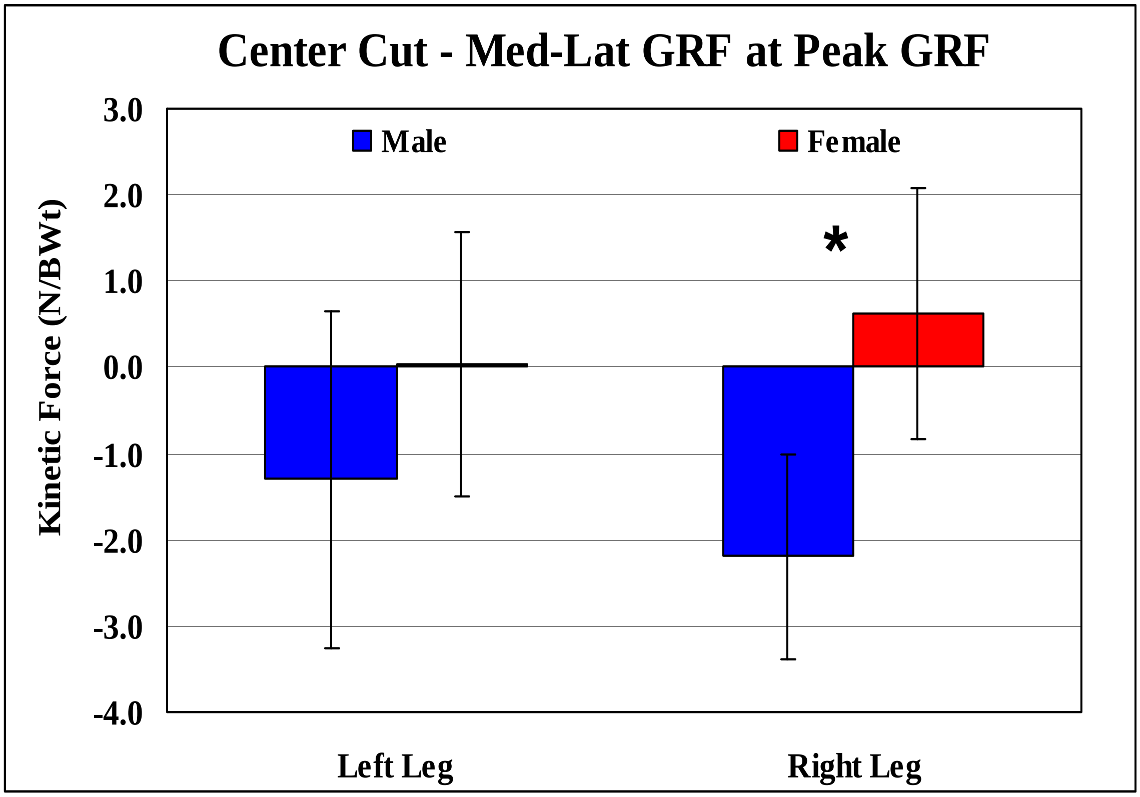


Figure 4.14. Statistically Significant Differences Medial-Lateral Ground Reaction Forces In The Right Leg While Performing A Straight Run With Positive Values Indicating Lateral And Negative Values Indicating Medial.

No statistical differences were noted between the genders for the kinetics of the left or right foot during initial contact (*Table 4.15*), peak knee flexion angles (*Table 4.16*), or push off (*Table 4.18*).

Right Cut

Statistical differences (p = 0.041, *d* = -0.80) were noted between the genders for the right foot ground reaction forces with the female participants sustaining greater anterior forces (0.23 N/kg ± 0.17 N/kg) than the male participants (0.02 N/kg ± 0.10 N/kg) during the initial contact (*Figure 4.15, Table 4.15*). No statistical differences were noted between the genders for the left foot ground reaction forces during initial contact (*Table 4.15*).

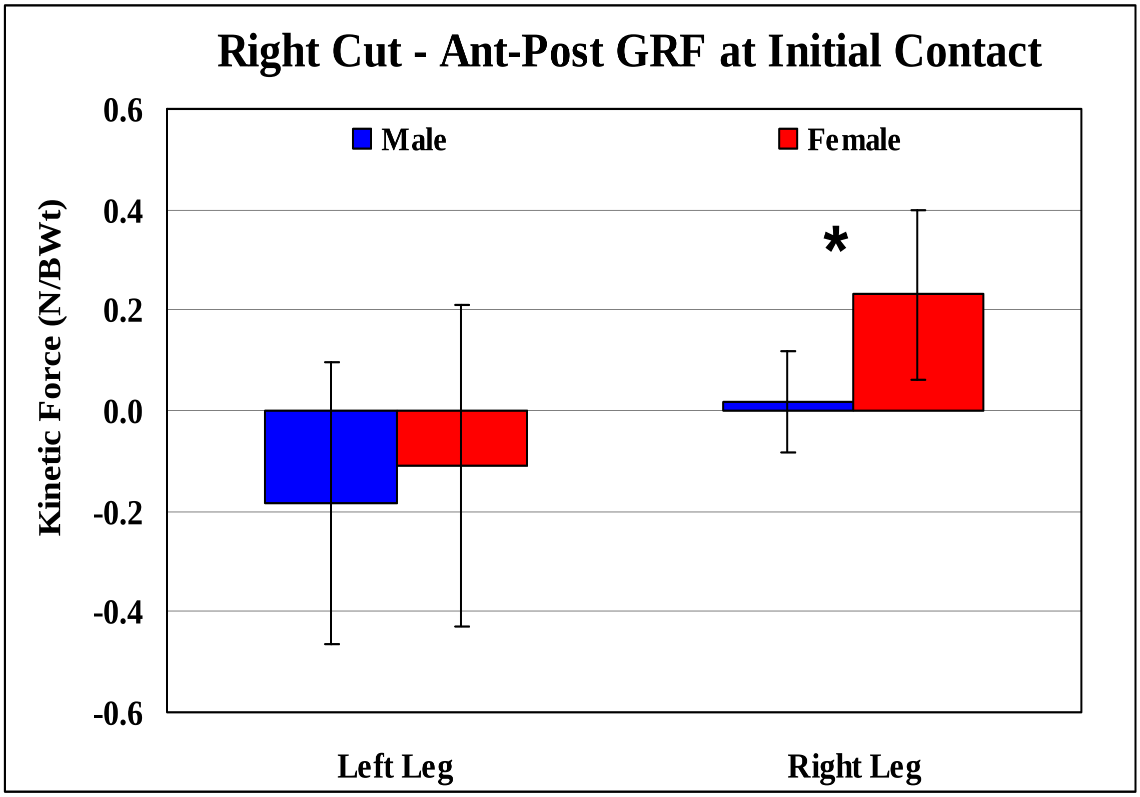


Figure 4.15. Statistically Significant Differences Anterior-Posterior Ground Reaction Forces In The Right Leg While Performing A Right Cut With Positive Values Indicating Anterior And Negative Values Indicating Posterior.

Statistical differences (p = 0.033, r = -0.84) were noted between the genders for the right foot ground reaction forces with the male subjects sustaining posterior forces (0.17 N/kg ± 0.61 N/kg) and female subjects sustaining anterior forces (1.19 N/kg ± 1.01 N/kg) during peak knee flexion angles (*Figure 4.16, Table 4.16*). No statistical differences were noted between the genders for the left foot ground reaction forces during peak knee flexion angles (*Table 4.16*).

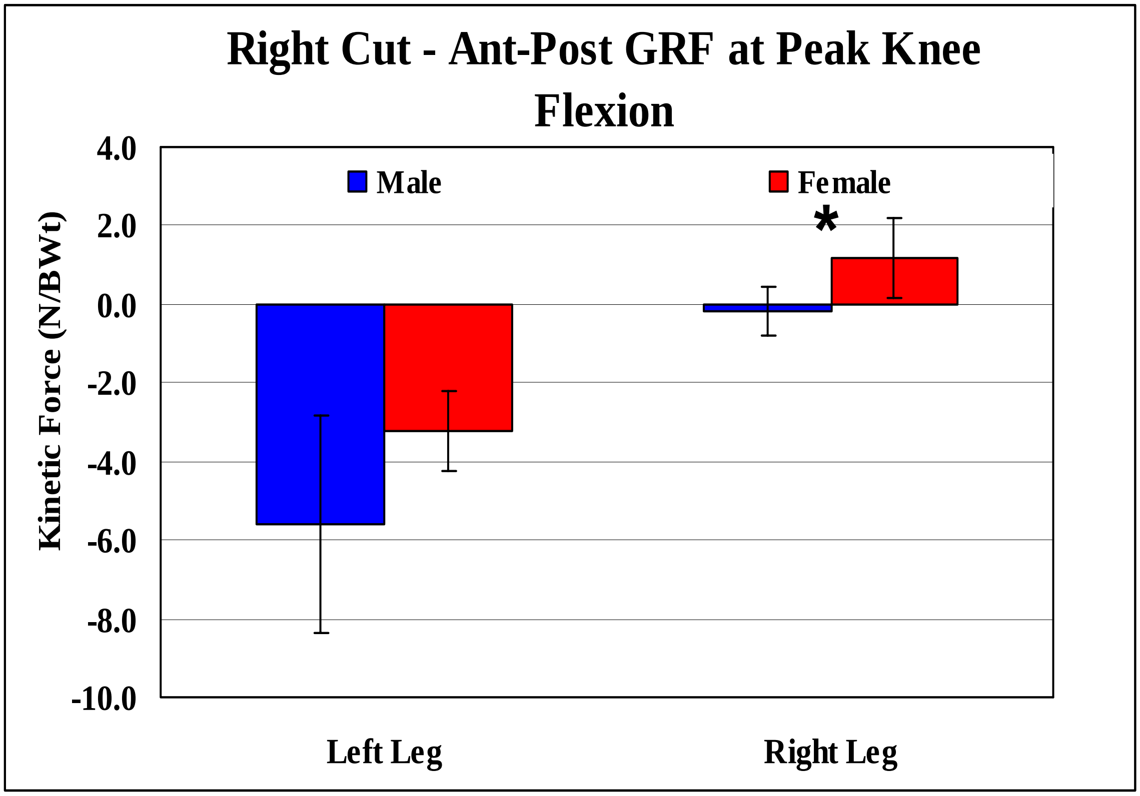


Figure 4.16. Statistically Significant Differences Anterior-Posterior Ground Reaction Forces In The Right Leg While Performing A Right Cut With Positive Values Indicating Anterior And Negative Values Indicating Posterior.

Statistical differences (p = 0.031, r = -0.89) were noted between the genders for the left foot ground reaction forces in the medial-lateral direction with male participants sustaining greater medial forces (3.22 N/kg ± 1.93 N/kg) than the female participants (0.76 N/kg ± 0.83 N/kg) during peak knee flexion angles (*Figure 4.17*, *Table 4.16*). No statistical differences were noted between the genders for the right foot ground reaction forces during peak knee flexion angles (*Table 4.16*).

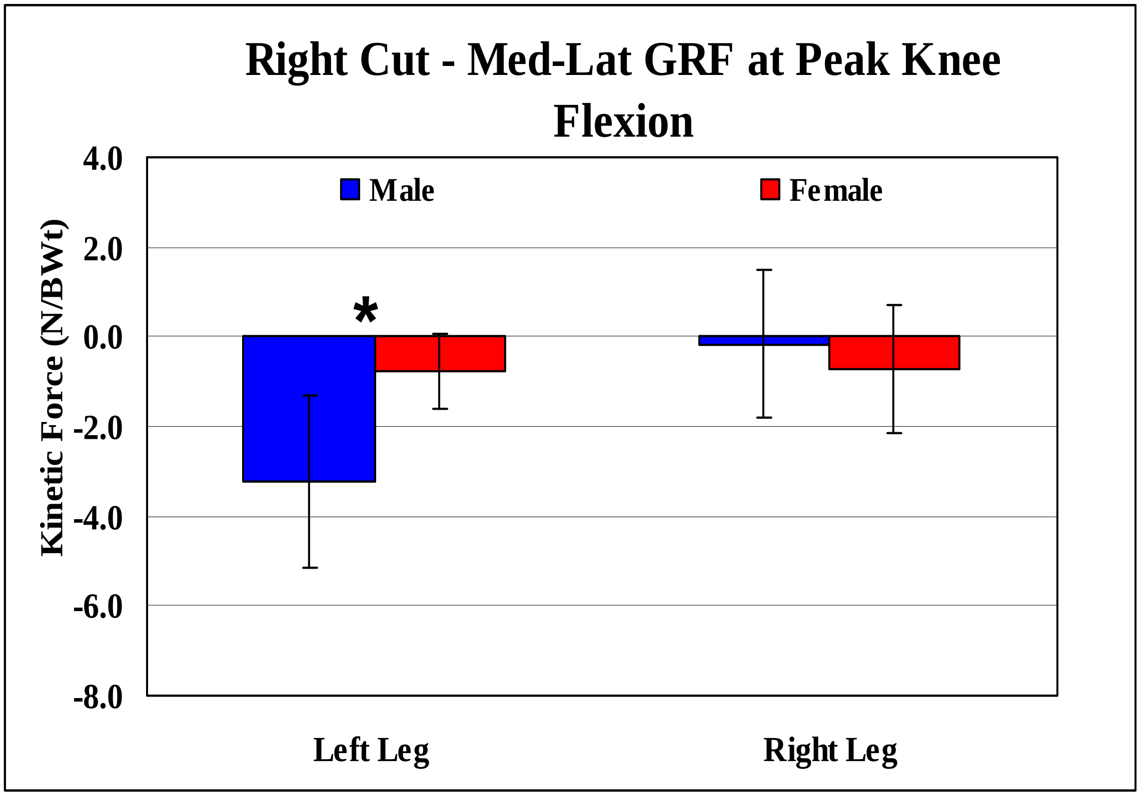


Figure 4.17. Statistically Significant Differences Medial-Lateral Ground Reaction Forces In The Right Leg While Performing A Right Cut With Positive Values Indicating Lateral And Negative Values Indicating Medial.

Statistical differences (p = 0.009, r = -1.12) were noted between the genders for the right foot ground reaction forces in the anterior-posterior direction with the male participants incurring posterior forces (0.36 N/kg ± 0.54 N/kg) and the female participants incurring anterior forces (1.30 N/kg ± 0.93 N/kg) during the peak GRF (*Figure 4.18, Table 4.17*). No statistical differences were noted between the genders for the left foot ground reaction forces during peak GRF (*Table 4.17*).

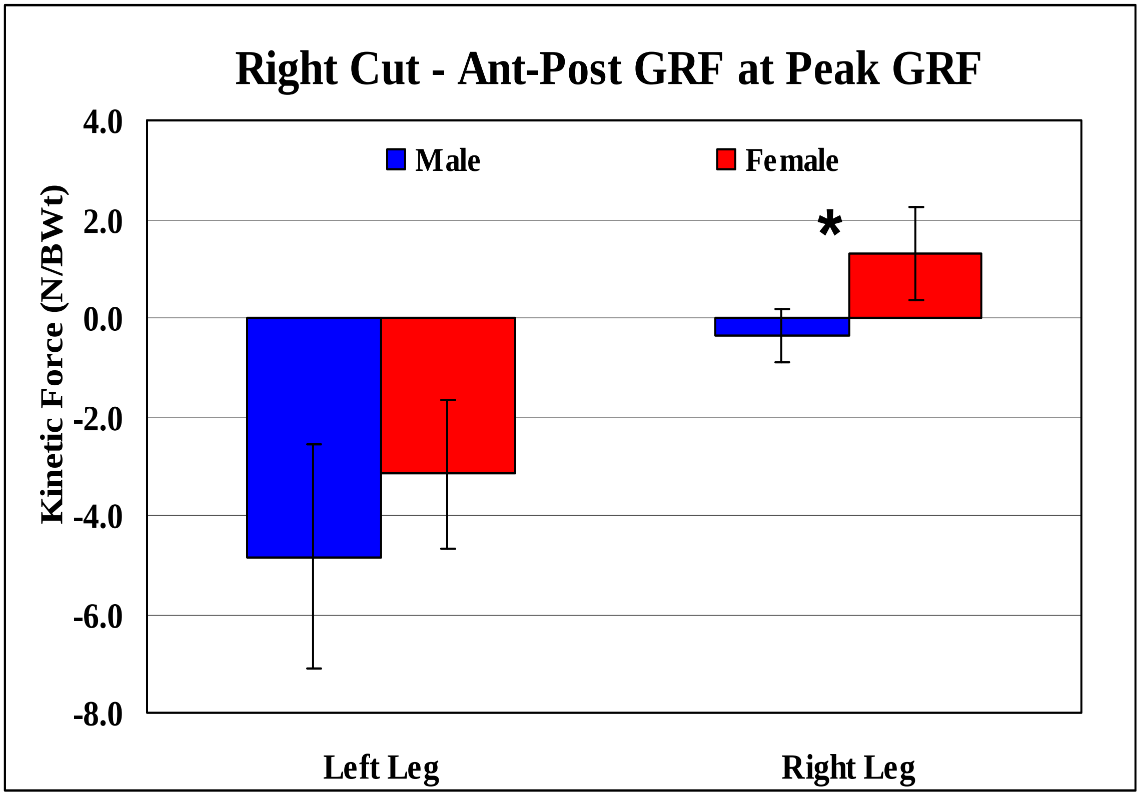


Figure 4.18. Statistically Significant Differences Anterior-Posterior Ground Reaction Forces In The Right Leg While Performing A Right Cut With Positive Values Indicating Anterior And Negative Values Indicating Posterior.

No statistical differences were noted between the genders for the kinetics of the left or right foot during push off (*Table 4.18*).

Table 4.15

Kinetic Forces Calculated During Initial Contact.

Initial Contact

			Left Ant-Post (N/BW)	Right Ant-Post (N/BW)	Left Med-Lat (N/BW)	Right Med-Lat (N/BW)	Left Vert (N/BW)	Right Vert (N/BW)
Left Cut	Male	Mean	0.10	0.14	0.33	0.35	1.35	1.22
		SD	0.13	0.20	0.41	0.41	0.80	1.11
	Female	Mean	-0.01	0.09	0.61	0.85	1.75	1.65
		SD	0.11	0.11	0.88	1.04	2.18	1.74
		T-Test	0.17	0.65	0.52	0.35	0.71	0.65
		SE	0.48	0.16	-0.23	-0.35	-0.14	-0.15
Center Cut	Male	Mean	0.07	0.05	0.30	0.16	1.33	0.95
		SD	0.22	0.21	0.38	0.15	0.69	0.51
	Female	Mean	-0.04	0.01	0.19	0.79	1.09	1.73
		SD	0.13	0.08	0.22	0.75	0.76	1.39
		T-Test	0.37	0.67	0.61	0.10	0.61	0.27
		SE	0.31	0.15	0.17	-0.71	0.17	-0.41
Right Cut	Male	Mean	-0.18	0.02	0.39	0.22	1.72	1.30
		SD	0.28	0.10	0.66	0.32	1.30	0.69
	Female	Mean	-0.11	0.23	0.68	1.09	1.89	2.20
		SD	0.32	0.17	1.01	0.93	2.18	1.51
		T-Test	0.70	0.04	0.61	0.08	0.89	0.26
		SE	-0.13	-0.80	-0.17	-0.70	-0.05	-0.41

🟨 = Stastical Significance ($p < 0.05$)

Table 4.16

Kinetic Forces Calculated During Peak Knee Flexion.

Peak Knee Flexion

			Left Ant-Post (N/BW)	Right Ant-Post (N/BW)	Left Med-Lat (N/BW)	Right Med-Lat (N/BW)	Left Vert (N/BW)	Right Vert (N/BW)
Left Cut	Male	Mean	0.64	4.93	-0.50	-2.37	4.34	15.92
		SD	0.58	3.32	0.69	3.02	4.91	11.55
	Female	Mean	-0.89	2.47	-0.26	-2.14	6.27	9.90
		SD	1.06	1.71	0.75	1.74	5.24	5.74
		T-Test	0.02	0.18	0.62	0.89	0.56	0.33
		SE	0.93	0.49	-0.16	-0.05	-0.19	0.35
Center Cut	Male	Mean	-1.10	0.97	-1.85	-1.82	9.90	10.30
		SD	1.14	0.94	1.63	1.71	8.29	8.49
	Female	Mean	-1.63	1.30	-0.81	-0.97	7.50	8.62
		SD	1.53	0.73	1.08	1.69	5.03	5.13
		T-Test	0.55	0.55	0.27	0.45	0.60	0.72
		SE	0.20	-0.20	-0.38	-0.25	0.18	0.12
Right Cut	Male	Mean	-5.59	-0.17	-3.22	-0.16	18.89	9.59
		SD	2.77	0.61	1.93	1.67	9.54	5.34
	Female	Mean	-3.24	1.19	-0.76	-0.71	12.65	10.41
		SD	1.01	1.01	0.83	1.43	2.85	3.59
		T-Test	0.11	0.03	0.03	0.59	0.20	0.78
		SE	-0.62	-0.84	-0.89	0.18	0.50	-0.09

🟨 = Stastical Significance ($p < 0.05$)

Table 4.17

Kinetic Forces Calculated During Peak GRF.

Peak GRF

			Left Ant-Post (N/BW)	Right Ant-Post (N/BW)	Left Med-Lat (N/BW)	Right Med-Lat (N/BW)	Left Vert (N/BW)	Right Vert (N/BW)
Left Cut	Male	Mean	2.04	4.74	-3.12	-0.91	11.04	23.40
		SD	3.75	2.57	5.95	3.44	9.89	13.11
	Female	Mean	-1.61	2.51	1.40	-0.57	10.52	15.06
		SD	0.82	0.85	1.19	2.68	2.83	1.12
		T-Test	0.07	0.10	0.13	0.86	0.91	0.19
		SE	0.80	0.65	-0.63	-0.06	0.04	0.59
Center Cut	Male	Mean	-1.65	1.10	-1.30	-2.19	17.50	16.82
		SD	0.99	0.79	1.95	1.19	12.30	6.02
	Female	Mean	-1.97	1.37	0.03	0.62	12.51	12.89
		SD	1.02	0.41	1.53	1.45	1.29	1.84
		T-Test	0.63	0.51	0.26	0.01	0.39	0.20
		SE	0.16	-0.23	-0.38	-1.06	0.37	0.50
Right Cut	Male	Mean	-4.85	-0.36	-1.57	-0.27	22.45	11.09
		SD	2.26	0.54	3.99	1.71	15.51	6.11
	Female	Mean	-3.15	1.29	-0.14	0.27	13.79	12.59
		SD	1.51	0.93	1.75	2.13	2.07	2.60
		T-Test	0.20	0.01	0.48	0.67	0.25	0.63
		SE	-0.45	-1.12	-0.25	-0.14	0.49	-0.17

🟨 = Stastical Significance ($p < 0.05$)

Table 4.18

Kinetic Forces Calculated During Push Off.

Push Off

			Left Ant-Post (N/BW)	Right Ant-Post (N/BW)	Left Med-Lat (N/BW)	Right Med-Lat (N/BW)	Left Vert (N/BW)	Right Vert (N/BW)
Left Cut	Male	Mean	0.53	3.24	-0.58	-2.76	3.69	9.51
		SD	0.58	3.18	0.65	2.41	5.22	8.40
	Female	Mean	-0.84	2.33	-0.31	-2.14	5.60	9.10
		SD	1.11	1.77	0.72	1.74	5.85	6.22
		T-Test	0.04	0.59	0.55	0.65	0.60	0.93
		SE	0.81	0.18	-0.20	-0.15	-0.17	0.03
Center Cut	Male	Mean	-0.52	0.78	-1.25	-1.77	5.47	6.45
		SD	0.85	0.84	1.20	1.92	4.74	5.51
	Female	Mean	-1.47	0.99	-0.88	-1.24	6.48	6.22
		SD	1.68	0.89	1.01	1.25	6.00	5.80
		T-Test	0.29	0.71	0.61	0.62	0.78	0.95
		SE	0.37	-0.12	-0.17	-0.17	-0.09	0.02
Right Cut	Male	Mean	-2.90	-0.44	-2.90	-0.40	8.89	3.69
		SD	2.94	0.41	2.50	0.79	7.87	3.92
	Female	Mean	-2.21	0.78	-0.92	-1.22	7.79	6.18
		SD	2.26	1.25	1.00	1.36	7.03	6.19
		T-Test	0.69	0.07	0.14	0.28	0.82	0.47
		SE	-0.13	-0.74	-0.57	0.38	0.07	-0.25

🟨 = Stastical Significance ($p < 0.05$)

Dominant Limb v. Non-Dominant Limb

This study also collected information from participants in regards to limb dominance during the jump, land and cutting maneuver. Participants were asked to identify a dominant limb (determined by asking participants which leg they would prefer to kick a soccer ball) during the study in the attempts to possibly account for other neuromuscular variables when analyzing the data. Out of the 10 participants only one female participant identified the left leg as the leg of preference when kicking a soccer ball, the other nine participants identified the right leg as the dominant extremity. It is possible that this one participant could be performing tasks differently than her fellow participants based on her dominant and non-dominant extremity; therefore, the data was evaluated by flipping the right and left means values for all dependent variables for this one particular participant so her dominant extremity would match up with the rest of the participant's dominant extremity.

In theory, if leg dominance is a determining factor in the execution of neuromuscular strategies in athletics, then we should see significant differences in the data set based on this one participant. If a person is right handed and is asked to perform a task with the right hand, it stands to reason they could be more familiar (more practiced or more skilled) with this given task constraint. If a person is asked to do the same task with the opposite extremity (in this case the left hand) it also stands to reason that the task could be constrained through extraneous variables (i.e., coordination, strength, manual dexterity, balance, etc…). By taking the mean values obtained during the right cut and switching them to the left cut, it would stand to reason, that we have possibly limited or accounted for one of the many confounding variables. The center cut was not changed or altered in any way as the participant was allowed to self select the neuromuscular strategy to accomplish a center cut (straight run initiated by limb of choice).

When the data compared right and left legs during athletic maneuver, several significant results were noted during the left and center cut for the kinematic variables, and all three cuts for the kinetic variables. None of the cut directions were significantly different for the EMG amplitude or Gluteus Medius muscle onset. When the left leg dominant female's data was transposed (switching the mean values for left and right cuts), we find very similar results. There are three variables which changed the level of significance (1 in the left cut and 2 in the right cut) although the trends remained constant across all variables.

In the left cut, the level of significance was moved to a significant value (from $p = 0.066$ to $p = 0.048$) in the left leg anterior-posterior forces in the Non-Dominant Leg during peak GRF. Males performed this part of the athletic maneuver with an anterior forces (2.04 ± 3.75) while females had a posterior force (2.15 ± 1.43).

In the right cut, two variables changed levels of significance moving from significant variables to values slightly above the a priori significance level of $p = 0.05$. The anterior forces demonstrated trends for females to sustain greater anterior forces ($p = 0.069$) in the female athletes (0.210 ± 0.182) compared to the male athletes (0.016 ± 0.097). The anterior-posterior forces ($p = 0.073$) demonstrated trends for males to sustain a posterior force (0.169 ± 0.614) while females sustained an anterior force (1.015 ± 1.126).

The following tables demonstrate the overall significance and trend data for the Left and Right comparison (*Table 4.19*) and for the Dominant and Non-Dominant comparison (Table *4.20*). The three grayed in boxes in the Dominant and Non-Dominant comparison demonstrate values which changed in levels of significance (either moving from not significant to levels of significance or moving from levels of significance to not significant). It should be noted that the values for the left leg during peak GRF that are noted during the Dominant and Non-Dominant Leg comparison are not present on the Right and Left Leg comparison because the values were not originally statistically significant in that analysis and therefore were not included.

Table 4.19

Right Leg Verses Left Leg Statistical Significance Comparison

Right v. Left Leg

	Left Cut	Center Cut	Right Cut
EMG	No Difference	No Difference	No Difference
KINEMATIC	Left Ankle ($p = 0.019$): Flexion-Extension (peak knee flexion) ♀ ↑ Dorsiflexion	Right Ankle ($p = 0.012$): Flexion-Extension (peak GRF) ♂ ↑ Dorsiflexion	No Difference
KINETIC	Left Leg ($p = 0.222$): Anterior-Posterior forces (peak knee flexion) ♂ Anterior ♀ Posterior	Right Leg ($p = 0.222$): Medial-Lateral forces (peak knee flexion) ♂ Medial ♀ Lateral	Right Leg ($p = 0.041$): Anterior-Posterior (initial contact) ♀ ↑ Anterior
	Left Leg ($p = 0.040$): Anterior-Posterior forces (push off) ♂ Anterior ♀ Posterior		Right Leg ($p = 0.041$): Anterior-Posterior forces (peak knee flexion) ♂ Posterior ♀ Anterior
			Left Leg ($p = 0.031$): Medial-Lateral (peak knee flexion) ♂ ↑ Medial
			Right Leg ($p = 0.040$): Anterior forces (peak knee flexion) ♂ Posterior ♀ Anterior

Table 4.20

Right Leg Verses Left Leg Statistical Significance Comparison

Dominant v. Non-Dominant Leg

	Left Cut	Center Cut	Right Cut
EMG	No Difference	No Difference	No Difference
KINEMATIC	Non-Dominant Ankle (p = 0.014): Flexion-Extension (peak knee flexion) ♀ ↑ Dorsiflexion	Dominant Ankle (p = 0.012): Flexion-Extension (peak GRF) ♂ ↑ Dorsiflexion	No Difference
KINETIC	Non-Dominant Leg (p = 0.013): Anterior-Posterior forces (peak knee flexion) ♂ Anterior ♀ Posterior	Dominant Leg (p = 0.222): Medial-Lateral forces (peak knee flexion) ♂ Medial ♀ Lateral	Dominant Leg (p = 0.069: Anterior-Posterior (initial contact)* ♀ ↑ Anterior
	Non-Dominant Leg (p = 0.040): Anterior-Posterior forces (push off) ♂ Anterior ♀ Posterior		Dominant Leg (p = 0.073): Anterior-Posterior forces (peak knee flexion)* ♂ Posterior ♀ Anterior
	Non-Dominant Leg (p = 0.048): Anterior-Posterior forces (peak GRF) ♂ Anterior ♀ Posterior		Non-Dominant Leg (p = 0.030): Medial-Lateral (peak knee flexion) ♂ ↑ Medial
			Dominant Leg (p = 0.008): Anterior forces (peak knee flexion) ♂ Posterior ♀ Anterior

*Approaching Significance ($p < 0.05$)

▨ = Change in Significance Level from Left v. Right

CHAPTER 5

Discussion & Clinical Relevance

Discussion

The amount of research dedicated to examining the effects of gender on the increase in magnitude of female anterior cruciate ligament injury is extensive; however, even with all of the attention researchers have devoted to the problem, we still do not fully understand the increased risk factors. Researchers have been able to suggest common mechanisms for the non-contact ACL injury including activities of acceleration, deceleration, jumping, landing and changes in direction (Decker et al., 2003; McLean et al., 2003; Moeller & Lamb, 1997; Slauterbeck et al., 2002; Toth & Cordasco, 2001).

Researchers have also explored different risk factors associated with the gender differences including issues in environmental, hormonal, anatomical, neuromuscular, or sport specific factors (Anderson et al., 2001; Moeller & Lamb, 1997; Cowley et al., 2006). Neuromuscular risk factors have been of particular interest because of the availability for modifications. If researchers could identify neuromuscular risk factors, then they could potentially create training protocols to decrease these determined differences among the genders and hopefully decrease the risk of ACL injury in the female athlete. Researchers have primarily focused on the kinetics and kinematics of the knee as the primary causative factor in the risk of ACL injuries; however, since the lower extremity functions as a closed kinetic chain, it is imperative that researchers begin to look at the ankle and hip in addition to the knee to determine the risk of knee injury in the female athletic population.

The purpose of this study was threefold: 1. To determine if significant discrepancies exist between the genders in electromyography (EMG) amplitude of the left and right gluteus medius; 2. To examine joint angles at the hip, knee, and ankle during the landing and unanticipated cutting for gender differences; and 3. To compare the ground reaction forces sustained by the participants during the landing and push off phases of this sport specific maneuver differed between the genders.

EMG

The muscles surrounding the hip, in particular the gluteus medius, provide structural support during the midstance of gait (Anderson & Pandy, 2003). The muscle also functions as the primary abductor of the hip and has been postulated to control internal rotation of the femur during locomotion (Hart et al., 2007). Decreases in gluteus medius onset and activation have been suggested to contribute to increases in hip adduction and internal rotation, resulting in the "position of no return" (Ireland, 1999).

The primary EMG finding of this investigation was that there were no differences in EMG amplitudes in the right and left gluteus medius between the genders. There was also no difference in muscle onset times for the right and left gluteus medius for the genders. Comparison of our data with findings in current literature is limited because I am unaware of any investigation in which the researchers evaluated the effects of gender on bilateral gluteal muscles during landing and cutting tasks.

Ambegaonkar et al. (2008) examined the muscle onset times for female basketball and dancing athletes during a drop jump and found no difference in the observed muscles between the groups. While this study only examined basketball players, male participants were also included. Our study also did not find significant differences in muscle onset between the groups.

Carcia & Martin (2007) examined gender differences before and after ground contact in the left and right gluteus medius during a drop jump and found no differences in the adult participants. Russell and colleagues (2006) also found no gender different between adult participant's dominant extremity gluteus medius EMG when performing a single leg drop jump. Hart et al. (2007) instructed male and female soccer players to perform a forward jump while they monitored the right gluteus medius. They suggested the average gluteus medius activity was significantly higher in the male participants than in the female participants. The study by Hart et al. (2007) utilized adult participants and only monitored a single lower extremity, therefore limiting the comparison between these studies.

Other studies have utilized cutting maneuvers (Colby et al. 2000), forward jump landings (Cowling & Steele, 2001), or drop landings (Rozzi et al. 1999) to compare neuromuscular force attenuation strategies between genders, however, none of these studies examined the effects at the right and left gluteus medius. Each of these study protocols and maneuvers could simulate forces and joint positions as experienced during athletic activity, more direct comparison is relatively difficult due to the subtle differences in neuromuscular landing strategies utilized during the research procedures.

This study used two different methods of normalizing the EMG data. We used traditional MVIC values and additionally normalized left and right runs with a straight ahead run. Running and cutting are dynamic activities and as such, we valued a dynamic normalization protocol. This was not the first time EMG data had been normalized using a dynamic protocol as Cowling & Steele (2001) used a similar protocol; however, studies since have heavily relied on the MVIC normalization method for examining EMG data. This research study examined different muscles including: rectus femoris, vastus medialis, vastus lateralis, semimembranosus, biceps femoris, and medial head of the gastrocnemius. Our study did not examine the same set of muscles, and therefore, we can make speculation but results cannot be directly compared between the two studies. More recently, Bolgia & Uhl (2007) utilized this method to examine surface EMG data for the gluteus medius during exercise. The researchers normalized the raw EMG results using the maximum voluntary isometric contraction (MVIC), mean dynamic, and peak dynamic activity EMG during standing, single leg, and side lying hip abduction exercises.

Bolgia & Uhl (2007) suggested the MVIC method had the greatest measurement reliability in the determination of muscle activation amplitudes between subjects in their study. Other authors suggest the dynamic normalization methods are more accurate (Burden, Trew, & Baltzopoulos, 2003; Winter & Yack, 1987; Yang & Winter, 1984). The exercises in Bolgia & Uhl’s research were not as dynamically challenging as an athletic maneuver, such as the ones performed in this study.

EMG Normalization

While none of the EMG normalized values were significant, it is extremely important to talk about them from a procedural standpoint. We normalized EMG values both to the maximum voluntary isometric contraction (MVIC) and we normalized the left and right cuts to a straight run (center cut). Very few studies have employed the dynamic normalization protocol. Most studies have utilized the MVIC normalization as the gold standard in EMG analysis; however, when we examine muscle contraction, we can see that even the basic mechanics is drastically different when performing an MVIC and a then a dynamic complex movement such as an athletic task. In a static MVIC, the person is not in motion and is asked to contract as hard as they can to perform a muscle contraction. There are several issues with this protocol. First and foremost, the level of effort given by the participant is extremely subjective. As an examiner, it is sometimes difficult to tell if a participant is giving a maximal effort. In a centralized dynamic movement, the participant is forced to give more of a maximal effort and it is usually a little bit easier to tell when someone is not performing a task as well or as fast as other given tasks.

In the data, there were two outliers which had to be removed from the center run normalization in the female participants. Both extreme values came from the same participant. There are several factors that could have altered the EMG signal obtained from this individual. Since the center run was a dynamic motion and the electrode was placed under clothing, there could be movement artifact between the electrode and skin, the clothing could have pulled on the electrode or rubbed up against it, or the participant could have hit the electrode during movement.

The results of this study, although not significant, suggest researchers should examine different forms of normalization which mimic the activity to be studied. Comparing a dynamic movement to another dynamic movement should help eliminate massive MVIC EMG percentages and help possibly to homogenize the data collection and analysis process. In addition, the added fact that participants are required to perform a task in which they can see tangible results (as opposed to contracting against an immovable object) should ensure that participants are giving a greater effort and help to increase the consistency of the normalized value. In the center run normalization, we could see the standard deviations drop compared to the MVIC normalization which should indicate that the center run EMG data collection was a better sample and more reproducable from trial to trial for each participant.

Kinematics

The primary kinematic findings of this investigation were that the male participants performed left cuts with less dorsiflexion in the left ankle (which is the lead leg for the cut) during peak knee flexion angles and that male participants performed center cuts with more dorsiflexion in the right ankle during peak ground reaction forces (GRFs). Our study did not find any statistical significance in any of the other joints as has been reported in previous research studies.

Our results do not concur with previous research which has indicated female participants perform athletic jumping or cutting maneuvers with less trunk flexion (DiStefano et al., 2005; Decker et al., 2003; McLean et al., 2004a; McLean et al., 2004b; Salci et al., 2004; Yu et al., 2006), less hip flexion (Decker et al., 2003; DiStefano et al., 2005; Ford et al., 2005; Jackson et al., 2008; Kernozek, et al., 2005; Kulas et al., 2008; McLean et al., 2004a; McLean et al., 2004b; Salci et al., 2004; Wikstrom et al., 2004; Yu et al., 2006), greater hip adduction (Hewett et al., 2005; Jackson et al., 2008; Jacobs et al., 2007, Pollard et al., 2004), greater hip internal rotation (Jackson et al., 2008; Pollard et al., 2004), greater knee abduction (Barber-Westin et al., 2005; Ferber et al., 2003; Ford et al., 2005; Jackson et al., 2008; Lephart et al., 2004, Pollard et al., 2004), and less knee flexion angles (Decker et al., 2003; DiStefano et al., 2005; McLean et al., 2004a; McLean et al., 2004b; Salci et al., 2004; Sell et al., 2006; Wikstrom et al., 2004; Yu et al., 2006) than their male counterparts. Many of these studies did not study adolescent populations, and some of them did not use athletic populations, therefore, this study is relatively unique.

In 2005, Ford and colleagues examined 126 middle and high school basketball athletes to examine jump-stop unanticipated cut maneuver. The participants were asked to perform a forward jump (0.4 m) and perform sidestep cuts in one of two directions (left or right 45°). The study suggests females performed tasks with greater knee abduction (valgus) angles when compared with male participants. The study also concluded that no differences were noted in the knee flexion angle at initial contact or maximum contact (peak ground reaction forces). While this study does not correspond with our results for the knee valgus angle, the population and task are very similar to our protocol. We did not find significance in the knee flexion angles in initial contact or peak ground reaction forces as well).

Brown and colleagues (2008) examined the effects of hip flexion and adduction angles during a single-leg landing. The study examined both male and female participants together and looked at differences in anticipated and unanticipated land and jumps. Brown et al. (2008) indicated participants had greater hip flexion and hip adduction angles at initial contact during the anticipated conditions than in unanticipated conditions. In addition, participants sustained less hip internal rotation at initial contact during anticipated conditions.

Though the task constraints of the protocol used in the study by Brown et al. (2008) were slightly different than those of this study, the results might help explain the lack of kinematic findings in this study. In conjunction with Brown et al. (2008), we were not able to distinguish any significance in hip flexion, hip abduction, or hip internal rotation during the different cutting maneuvers either. The maneuvers our participants were asked to complete were all unanticipated cut tasks therefore, it appears the neuromuscular control of the individual athlete's change response in conjunction with the amount of time available to perform tasks.

One of our original hypotheses stated female athletes would perform tasks with less hip and knee flexion, thus landing in a more extended position. Kernozek et al. (2005) examined kinematic differences in the frontal and sagittal plane during drop landings in adult male and female recreational athletes. The study suggested females performed tasks with greater hip flexion, knee flexion, and ankle dorsiflexion angles in the sagittal plane and greater peak knee valgus and ankle pronation in the frontal plane. This study did not find any differences in these values other than the dorsiflexion angles in the left and center cut. The variance in the participant pool could be attributed to the differences in statistical findings.

The kinematic joint angles of the left ankle (p = 0.019) during peak knee flexion angles of the left cut demonstrated that females performed athletic tasks with greater dorsiflexion than the male participants. However, in the right ankle, males had more dorsiflexion than female participants during the peak ground reaction force (GRF) (p = 0.012). The participants were using the left leg as a lead leg during the left cut and therefore appeared to be planning the cutting maneuver with greater detail than their male participants. The females spent more time in contact with the force plate than did the males which could suggest it took the females long time to decelerate from the land and then begin the propulsive forces to initiate the cut. The males on the other hand, performed these tasks quite rapidly and therefore might not have incurred as much dorsiflexion during the peak knee flexion. The participants were allowed to use either the right or left leg as a lead leg in the center cut (straight run) and therefore, it is difficult to distinguish exactly why men were performing tasks with an increase in the dorsiflexion angle. The male participants were more likely to initiate the center cut with the right leg (56.25%) compared to the females (33.33%) which could indicate a predisposition for males to utilize the Dominant limb as a lead leg and the Non-Dominant limb as a stance or plant leg. This position would put them in a very similar position for identifying the Dominant extremity (standing on the Non-Dominant Limb while kicking a soccer ball with the Dominant Limb). The females displayed the opposite result, with the majority of the center cuts being initiated by the left leg as a lead leg and the right leg being used as the stance leg.

Gender differences in reaction time have been reported in the literature with the females reacting to stimuli slower than their male counterparts (Adams et al., 1999;

Nobel et al., 1964). Nobel, Baker, & Jones (1964) discovered males had significantly faster reaction times than did females. It appears age could also play a role in reaction time. Williams and colleagues (2005) suggested reaction time fits on a U-shaped curve based on age and consistency. The younger we are and the older we are, the more inconsistent our decisions and the longer the reaction time. Our participants were in the adolescent age population and are on the up swing of this U-shaped curve. Hogan and colleagues (2005) found that adolescents had less errors and responded to stimuli sooner than their adult counterparts.

According to Adams et al. (1999), choice reaction time may not be the same between men and women. College-age male participants had an advantage in reaction time over their female counterparts. In addition, males also saw time advantages when the number of outcomes was multiplied. Hongwei and colleages (2006) found similar results between the genders for a test on college students. The researchers used EMG to test the visual-manual choice reaction time by monitoring the pronator teres and biceps brachii.

Blough & Slavin (1987) also proposed gender differences in reaction time and found that while women were more accurate in their decision making during tasks where they had to make a choice between several stimuli; however, they were also significantly slower than their male counterparts. The results from this study might have a significant impact on the current study as the female participants appeared to be landing with more dorsiflexion (except in the straight cut) and then taking a long time to initiate the cutting motion than the male participants.

Kinetics

The ACL injury mechanism is commonly described as deceleration, changes in direction, landing from a jump, or knee hyperextension. The primary restraint of the anterior cruciate ligament is to prevent anterior tibial translation. If participants are subjected to large anterior forces during landing and cutting, the ACL could be at risk for injury.

The primary kinetic findings of this investigation were that: 1) male participants sustained greater anterior forces in the left leg during peak knee flexion angles and push off during the left cut; 2) male participants sustained medial forces and female participants sustained lateral forces during the center cut; 3) female participants sustained greater anterior forces in the right leg than males during the peak flexion angles in the right cut; 4) the females sustained anterior forces, while males sustained posterior forces in the right leg during the peak ground reaction forces (GRF) in the right cut; and 5) the male participants sustained greater medial forces in the left foot during the peak knee flexion angles while cutting to the right.

Quatman et al. (2006) examined the effect of maturation on the gender differences in landing forces and performance of vertical jumps. The study indicated as male participants matured, the height of vertical jump increased; however the same was not the case for female participants. In addition, male participants decreased landing ground reaction forces, but girls did not. While our study did not examine large variations in maturation, all of our participants were between the ages of 13-17 years old; we did see a trend of larger ground reaction forces in the female athletes.

Significant differences were noted between genders during peak ground reaction forces in the center run and side cut in adolescent youth soccer players (Sabick et al., 2008). The side cut in this study required the athlete to land with each foot on a force plate and then lead with the right foot to attempt a side-step cut. The data from this study demonstrates similar results with significant forces in the center run and right cut during peak ground reaction forces. The center run was statistically significant in the medial-lateral direction, while the right cut was statistically significant in the anterior-posterior direction.

The landing forces in the anterior direction were similar to a previous study undertaken in the laboratory. Boham (2007) examined the effects of fatigue on the ground reaction forces for female collegiate soccer athletes during unanticipated cutting and discovered fatigue induced the increase in anterior and medial-lateral forces. While this study does not include a fatiguing protocol, it appears the differences the force directions of significance remain consistent. The primary function of the ACL is to restrain anterior tibial translation and the secondary function of the ACL is to resist valgus/varus (medial-lateral) loading during locomotion. With significant ground reaction forces occurring in these directions, further research is need to examine the effects of these forces on the injury mechanism (Boham, 2007; Boham, Harris, Pfeiffer, DeBeliso, & Kuhlman, 2008).

Sell et al. (2006) focused on landing tasks during planned and unplanned jumps in various directions (left, right, or vertical jumping) and determined the jumps to the left had greater vertical and posterior ground reaction forces compared with right and vertical jumps. The vertical jumps also demonstrated significantly greater vertical and posterior ground reaction forces than the right cut. Our data collection did not follow these trends and did not find any statistical significance for the vertical or posterior ground reaction forces during any of the cutting directions. Sell et al. (2006) only examined the right leg during testing; therefore, this study could offer an alternative explanation for the force productions occurring in the dominant and non-dominant lower extremities in adolescent athletes during athletic maneuvers.

The female athletes in this study appeared to have significantly different landing force accommodation than did the adolescent male athletes. Cuts to the right appeared to have the greatest significance for the right leg with females sustaining greater anterior forces and for the left leg with the males sustaining greater medial forces. In the left cut, males appeared to have greater anterior forces in the left leg. The center cut had males sustaining medial forces while females sustained lateral forces in the right leg. The center cut is the only cut in which the forces were highest in the plant leg (trail leg). The medial forces seen in the center cut for the males could have indicated the males were landing and taking off with more precision than were the females who appeared to distribute their weight to the lateral portion of the foot prior to initiating the cut.

The differences in kinetic forces could indicate differences in the landing protocol employed by male and female adolescent athletes during unanticipated cutting tasks. Differences in forces could indicate risk for ACL injury. The anterior forces are of particular concern as the primary restrain of the ACL is to resist these anterior forces during athletic maneuvers.

Dominant Limb v. Non-Dominant Limb

While the data from this sample did not change dramatically with the Dominant and Non-Dominant leg comparison, this still could prove to be an important factor in the increased rate of injury in the female athlete. It is difficult to determine the effects of limb dominance on the decision making and neuromuscular strategy of athletes. This study has some significant limitations in the analysis of the statistical measures. The sample population was very small; however, even with these few of numbers we were still able to obtain some levels of significance. Therefore, I believe accounting for limb dominance in future measures of athletic activity might provide some useful information giving us further insight into the ACL injury paradigm.

Clinical Relevance

This study takes a novel approach to research, as it is one of very few studies to examine both the left and right lower extremity during dynamic athletic maneuvers and to examine the effects of gender on an adolescent athletic population. This study does not illustrate dramatic differences between the genders when examining potential muscle activation strategies of the gluteus medius during unanticipated athletic cutting activities. With future studies, researchers should continue to examine the differences in EMG strategies and joint kinematics to determine if pre-participation physical examinations of the bilateral strength of the hips in adolescent athletes. Although the results of this study did not demonstrate significant differences between the genders, researchers should continue to educate and examine training protocols, which enhance the strength of the musculature of the hips in the attempt to prevent non-contact ACL injuries in the adolescent female athlete.

Future research should examine the effects of ACL prevention training protocols and attempt to further examine the effects of the significant force differences seen between the genders. In addition, athletes from other sports should be examined using similar protocols to examine for sport specific differences between and among adolescent athletes. Fatigue is also a very important factor in the neuromuscular strategies of athletes and as such should be examined to determine if differences occur between the genders.

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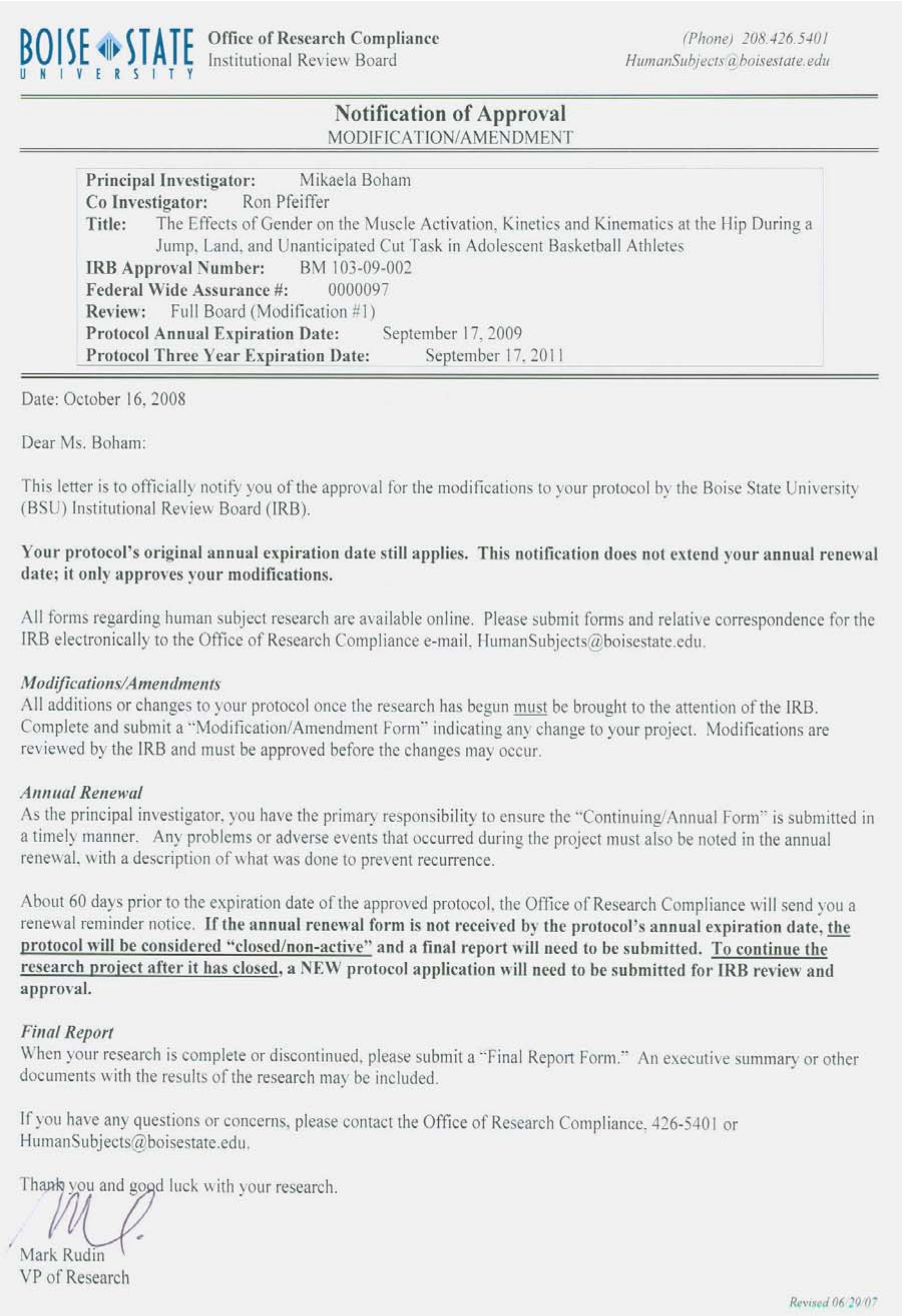
APPENDIX A

Notification of Dissertation Approval



BOISE STATE UNIVERSITY
Office of Research Compliance
Institutional Review Board

(Phone) 208.426.5401
HumanSubjects@boisestate.edu

Notification of Approval

MODIFICATION/AMENDMENT

Principal Investigator: Mikaela Boham
Co Investigator: Ron Pfeiffer
Title: The Effects of Gender on the Muscle Activation, Kinetics and Kinematics at the Hip During a Jump, Land, and Unanticipated Cut Task in Adolescent Basketball Athletes
IRB Approval Number: BM 103-09-002
Federal Wide Assurance #: 0000097
Review: Full Board (Modification #1)
Protocol Annual Expiration Date: September 17, 2009
Protocol Three Year Expiration Date: September 17, 2011

Date: October 16, 2008

Dear Ms. Boham:

This letter is to officially notify you of the approval for the modifications to your protocol by the Boise State University (BSU) Institutional Review Board (IRB).

Your protocol's original annual expiration date still applies. This notification does not extend your annual renewal date; it only approves your modifications.

All forms regarding human subject research are available online. Please submit forms and relative correspondence for the IRB electronically to the Office of Research Compliance e-mail, HumanSubjects@boisestate.edu.

Modifications/Amendments
All additions or changes to your protocol once the research has begun must be brought to the attention of the IRB. Complete and submit a "Modification/Amendment Form" indicating any change to your project. Modifications are reviewed by the IRB and must be approved before the changes may occur.

Annual Renewal
As the principal investigator, you have the primary responsibility to ensure the "Continuing/Annual Form" is submitted in a timely manner. Any problems or adverse events that occurred during the project must also be noted in the annual renewal, with a description of what was done to prevent recurrence.

About 60 days prior to the expiration date of the approved protocol, the Office of Research Compliance will send you a renewal reminder notice. **If the annual renewal form is not received by the protocol's annual expiration date, the protocol will be considered "closed/non-active" and a final report will need to be submitted. To continue the research project after it has closed, a NEW protocol application will need to be submitted for IRB review and approval.**

Final Report
When your research is complete or discontinued, please submit a "Final Report Form." An executive summary or other documents with the results of the research may be included.

If you have any questions or concerns, please contact the Office of Research Compliance, 426-5401 or HumanSubjects@boisestate.edu.

Thank you and good luck with your research.

Mark Rudin
VP of Research

Revised 06/29/07

APPENDIX B

Informed Assent Form

AND

Informed Consent Form

Boise State University
Consent to be a research participant
Minor Assent Form

PURPOSE AND BACKGROUND

Mikaela Boham, Graduate Assistant and doctoral student in the Department of Kinesiology and College of Education – Curriculum, Instruction, and Foundational Studies is conducting a research study to determine if gender alters the kinematics and kinetics of a jump, land, and unanticipated side cut in adolescent basketball players. I am being asked to participate in this study because I am member of a club basketball team in the Boise area between the ages of 13 and 16 years of age.

PROCEDURES

If I agree to be in the study, the following will occur. I will participate in a single testing session with multiple components.

1. I will be asked to wear my own athletic clothing for the testing session. Clothing needs to be snug on the body to prevent clothing movement during exercise. Females will be asked to wear either a one piece swimming suit (spandex shorts may be worn over the bottom of the suit), or spandex shorts and a sports bra or dry fit shirt. Males will be asked to wear spandex shorts and either a dry fit shirt or no shirt. For both males and females, baggy shorts may be worn over the spandex; however, we will be asking people to tuck the bottom of the shorts into the spandex so we can attach makers for data collection.
2. I will be asked to report to the Intermountain Orthopaedic Sports Medicine and Biomechanics Research Laboratory located within the Micron Engineering Complex (Room 105).
3. I will then be asked to read and sign an informed consent (adult accompanying the minor) and informed assent (minor) and be informed of my rights as a research participant.
4. I will be asked for demographic information by the researcher. I will be asked to provide information about: age (date of birth), height, weight, gender, and years of playing experience. For this research project, we are requesting demographic information. Due to the make-up of Idaho's population, the combined answers to these questions may make an individual person identifiable. We will make every effort to protect participants' confidentiality. However, if you are uncomfortable answering any of these questions, you may leave them blank.
5. I will warm-up with light calisthenics, dynamic stretching, and traditional speed drills (fast, low intensity, skips and hops) at the biomechanics lab. Immediately following warm up a same sex research assistant will prepared my skin for placement of electromyography (EMG) electrodes to determine the actions of my muscles under the skin. The skin will be shaved to remove hair, then rubbed with

an alcohol swab and slightly abraded with a pumice stone to remove access dead skin and create good conduction for collection of EMG data.

6. I will be fitted on the Biodex II isokinetic machine to test my hip strength. I understand I will be lying on an isokinetic machine with restraints across my torso (around my iliac crest [hip bone]). A standard knee attachment device will be secured to the leg so the pad is placed between the knee and the hip. I will be required to perform 5 isometric contractions (contracting the muscles without moving the joint) to assess the maximum strength for the hip flexion, hip extension, hip adduction, and hip adduction.
7. I will then be fitted with reflective markers for motion capturing analysis. The reflective markers will be attached to my body using double sided tape and/or elastic bands.
8. I will be asked to perform 3 maximum vertical jumps to calculate a vertical jump height. The researcher will calculate 50% of my average maximum vertical jump height and place a hurdle at that height in front of two ground reaction force plates.
9. I will be asked to perform 15 jump, land, and unanticipated cut assessments consisting of a forward jump from a line taped on the floor over a hurdle and land on a ground level force plate (jump distance will be approximately 120-150 cm), I will then be shown a light that will prompt a cut in one of three directions (45° to the right, straight ahead, and 45° to the left). I will complete a total of 15 jumps (5 in each direction presented in a randomized order).

I will be allowed to terminate any or all activity at any time if I choose to with no questions from the researcher. As a minor, my parent and/or legal guardian will be allowed to terminate any or all activity with no questions from the researcher. I will immediately cease activity if I feel light headed, nauseous, or ill in anyway.

The procedures will take approximately two hours to complete the single session. The student’s parent or legal guardian will be present for all testing.

RISKS/DISCOMFORTS

The physical activity in this study could result in muscle and tendon injury and/or some soreness, abnormal blood pressure, fainting, disorders of heartbeat, and in rare instances heart attack, stroke, or death. However, the risks of cardiovascular complications in this population are less than one per one million person hours of activity. In the case of any of these rare instances occuring, standard emergency procedures will be followed.

Confidentiality: Participation in research may involve a loss of privacy; however, my records will be handled as confidentially as possible. Only members of the lab staff will have access to my study records. After all the data has been analyzed, it will be archived on a computer. No individual identities will be used in any reports or publications that may result from this study.

I am aware that choosing to participate in this study might place me at an additional risk for an injury. As a result of a serious injury, I might miss practice participation or even face the loss of an athletic scholarship.

BENEFITS

There will be no direct benefit to me from participating in this study. However, information gained from the testing in this study will hopefully shed light on the ongoing and perplexing problem concerning the excessive numbers of lower extremity, in particular ACL injuries, to the female athletic population.

COSTS

There will be no costs to me as a result of taking part in this study.

This University is not able to offer financial compensation nor to absorb the costs of medical treatment should you be injured as a result of participating in this research.

PAYMENT

I will not be paid for my participation in this study.

QUESTIONS

I have talked to Miss Mikaela Boham or Dr. Ron Pfeiffer about this study and have had my questions answered. If I have further questions, I may call them at (208) 426-5710 (Biomechanics Lab), (208) 426-3709 (Dr. Ron Pfeiffer's office) or at (208) 426-1053 (Mikaela Boham).

If I have any comments or concerns about participation in this study, I should first talk with the investigator. If for some reason I do not wish to talk to the research investigator, I may contact the Institutional Review Board, which is concerned with the protection of volunteers in research projects. I may reach the board office between 8:00 AM and 5:00 PM, Monday through Friday, by calling (208) 426-1574 or by writing: Institutional Review Board, Office of Research Administration, Boise State University, 1910 University Dr., Boise, ID 83725-1135.

ASSENT

I will be given a copy of this assent form to keep.

Participation in research is voluntary. I am free to decline to be in this study, or to withdraw from it at any point. My decision as to whether or not to participate in this study will have no influence on my present status as a student athlete.

If I agree to participate I should sign below.

______________________________ ________________
Signature of Study Participant Date

______________________________ ________________
Signature of Person Obtaining Consent Date

I am allowing the researcher to use 2-Dimension Digital Video during the data collection. I have the right to refuse video collection during data collection and still participate in the study. If the researcher uses any information, they will block out any identifiable features of mine.

______________________________ ________________
Signature of Study Participant Date

This project has been reviewed by the Boise State University Institutional Review Board for the Protection of Human Participants in Research IRB# BM 103-09-002 (208-426-1574).

Boise State University

Consent to be a research participant

Parental Consent Form

PURPOSE AND BACKGROUND

Mikaela Boham, Graduate Assistant and doctoral student in the Department of Kinesiology and College of Education – Curriculum, Instruction, and Foundational Studies is conducting a research study to determine if gender alters the kinematics and kinetics of a jump, land, and unanticipated side cut in adolescent basketball players. I am being asked to participate in this study because I am member of a club basketball team in the Boise area between the ages of 13 and 16 years of age.

PROCEDURES

If I agree to let my son or daughter participate in this study, the following will occur. My son or daughter will participate in a single testing session with multiple components.

10. My son or daughter will be asked to report to the Intermountain Orthopaedic Sports Medicine and Biomechanics Research Laboratory located within the Micron Engineering Complex (Room 105). My son or daughter will be asked to read and sign an informed assent and I will be informed of my rights as a research participant. I will be asked to read and sign an informed consent (adult accompanying the minor) and I will be informed of my child's rights as a research participant.
11. My son or daughter will be asked for demographic information by the researcher. I will be asked to provide information about: age (date of birth), height, weight, gender, and years of playing experience. For this research project, we are requesting demographic information. Due to the make-up of Idaho's population, the combined answers to these questions may make an individual person identifiable. We will make every effort to protect participants' confidentiality. However, if you are uncomfortable with your child answering any of these questions, he or she may leave them blank.
12. My son or daughter will warm-up with light calisthenics, dynamic stretching, and traditional speed drills (fast, low intensity, skips and hops) at the biomechanics lab. Immediately following warm up a same sex research assistant will prepared my son or daughter's skin for placement of electromyography (EMG) electrodes to determine the actions of my muscles under the skin. The skin will be shaved to remove hair, then rubbed with an alcohol swab and slightly abraded with a pumice stone to remove access dead skin and create good conduction for collection of EMG data.
13. My son or daughter will be fitted on the Biodex II isokinetic machine to test for hip strength. My son or daughter will be lying on an isokinetic machine with restraints across his or her torso (around the iliac crest [hip bone]). A standard knee attachment device will be secured to the leg so the pad is placed between the knee and the hip. My son or daughter will be required to perform 5 isometric

contractions (contracting the muscles without moving the joint) to assess the maximum strength for the hip flexion, hip extension, hip adduction, and hip adduction.

14. My son or daughter will then be fitted with reflective markers for motion capturing analysis. The reflective markers will be attached to the body using double sided tape and/or elastic bands.
15. My son or daughter will be asked to perform 3 maximum vertical jumps to calculate a vertical jump height. The researcher will calculate 50% of my son or daughter's average maximum vertical jump height and place a hurdle at that height in front of two ground reaction force plates.
16. My son or daughter will be asked to perform 15 jump, land, and unanticipated cut assessments consisting of a forward jump from a line taped on the floor over a hurdle and land on a ground level force plate (jump distance will be approximately 120-150 cm). My son or daughter will then be shown a light that will prompt a cut in one of three directions (45° to the right, straight ahead, and 45° to the left). My son or daughter will complete a total of 15 jumps (5 in each direction presented in a randomized order).

My son or daughter will be allowed to terminate any or all activity at any time if I choose to with no questions from the researcher. As a parent and/or legal guardian, I will be allowed to terminate any or all activity with no questions from the researcher.

The procedures will take approximately two hours to complete the single session. As the student's parent or legal guardian I will be present for all testing.

RISKS/DISCOMFORTS

The physical activity in this study could result in muscle and tendon injury and/or some soreness, abnormal blood pressure, fainting, disorders of heartbeat, and in rare instances heart attack, stroke, or death. However, the risks of cardiovascular complications in this population are less than one per one million person hours of activity. In the case of any of these rare instances occuring, standard emergency procedures will be followed.

Confidentiality: Participation in research may involve a loss of privacy; however, my records will be handled as confidentially as possible. Only members of the lab staff will have access to my study records. After all the data has been analyzed, it will be archived on a computer. No individual identities will be used in any reports or publications resulting from this study.

I am aware choosing to participate in this study might place me at an additional risk for an injury. As a result of a serious injury, I might miss practice participation, competitions or even face the loss of an athletic scholarship.

BENEFITS

There will be no direct benefit to me or my son or daughter from participating in this study. However, information gained from the testing in this study will hopefully shed light on the ongoing and perplexing problem concerning the excessive numbers of lower extremity, in particular ACL injuries, to the female athletic population.

COSTS

There will be no costs to me as a result of taking part in this study.

This University is not able to offer financial compensation nor to absorb the costs of medical treatment should the participant be injured as a result of participating in this research.

PAYMENT

I will not be paid for my son or daughter's participation in this study.

QUESTIONS

I have talked to Miss Mikaela Boham or Dr. Ron Pfeiffer about this study and have had my questions answered. If I have further questions, I may call them at (208) 426-5710 (Biomechanics Lab), (208) 426-3709 (Dr. Ron Pfeiffer's office) or at (208) 426-1053 (Mikaela Boham).

If I have any comments or concerns about participation in this study, I should first talk with the investigator. If for some reason I do not wish to talk to the research investigator, I may contact the Institutional Review Board, which is concerned with the protection of volunteers in research projects. I may reach the board office between 8:00 AM and 5:00 PM, Monday through Friday, by calling (208) 426-1574 or by writing: Institutional Review Board, Office of Research Administration, Boise State University, 1910 University Dr., Boise, ID 83725-1135.

CONSENT

I will be given a copy of this consent form to keep.

Participation in research is voluntary. I am free to decline to be in this study, or to withdraw from it at any point.

If I agree to allow my son or daughter's participation I should sign below.

Signature of Study Participant's Parent or Legal Guardian Date

Signature of Person Obtaining Consent Date

I agree to allow the researcher to use 2-Dimension Digital Video during the data collection on my son or daughter. We have the right to refuse video collection during data collection and still participate in the study. If the researcher uses any information, they will block out any identifiable features of my son or daughter.

Signature of Study Participant Date

This project has been reviewed by the Boise State University Institutional Review Board for the Protection of Human Participants in Research IRB# BM 103-09-002 (208-426-1574).

APPENDIX C

Warm-Up Protocol

WARM-UP PROTOCOL

The warm-up will be the same for all participants in the study. The warm-up will be conducted in the biomechanics lab or on the adjacent grassy structure. The warm-up will consist of light callisthenic activity such as jogging, and dynamic stretching (slow high knees, slow squats, grapevine, etc.), and some traditional speed warm-up drills (fast, low intensity skips and hops). The warm-up will be concluded with a light stretching procedure to reduce the risk of injury during participation.

1. Warm-up stretching, quick lower extremity stretch (gastrocnemicus, hamstring, quadriceps).

2. 25 yard slow high knees

3. 25 yard jogging, 1-2 @ 50% effort, 1-2 @75% effort, 1-2 @ 100% effort

4. 25 yard high knees, 4-5 repetitions

5. 25 yard grapevine, 4-5 repetitions

6. 25 yard skips, 4-5 repetitions

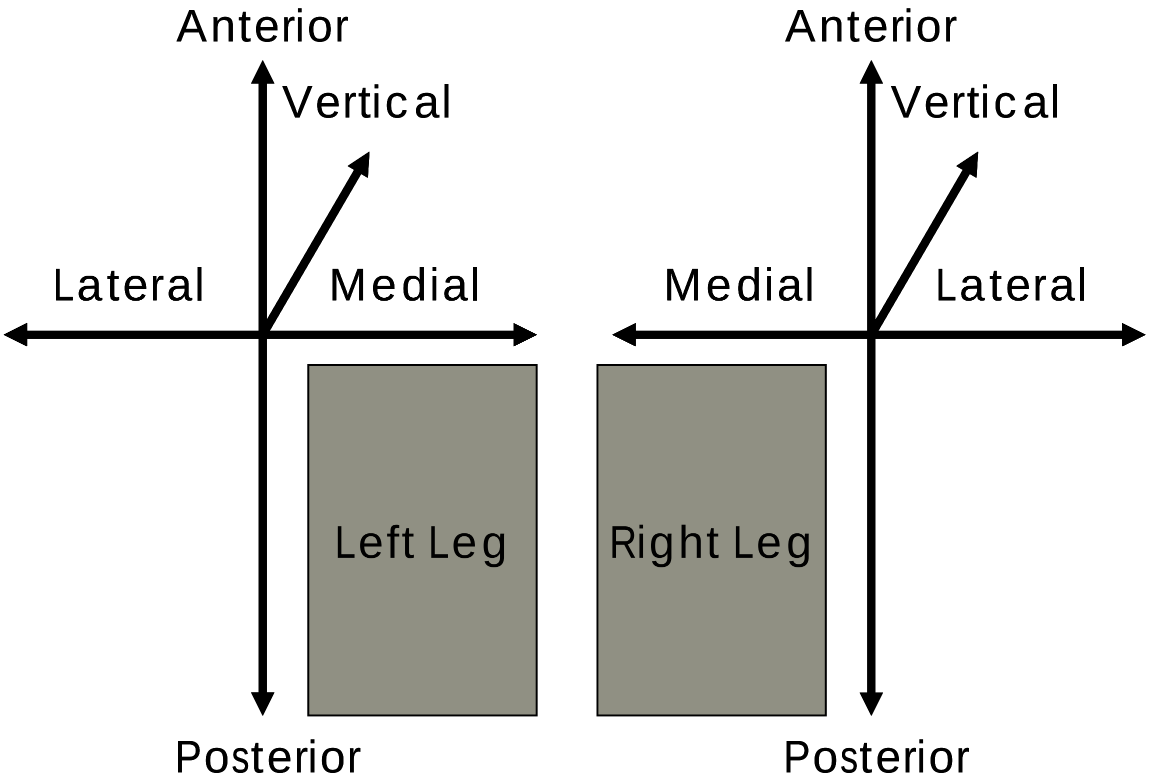
7. Full body stretching (gastrocnemius, hamstring, quadriceps, hip adductors, hip flexors, low back stretch, shoulder stretch, arm stretch, neck stretches)

APPENDIX D

Ground Force Plate

Global Coordinate System

GROUND FORCE PLATE
GLOBAL COORDINANT SYSTEM



X Forces – medial and lateral components
Y Forces – anterior and posterior components
Z Forces – vertical ground components

Each leg (right and left) has a global coordinate system. The force plates were oriented so the X-axis force values coming from the lateral left foot or lateral right foot are positive in force value, and forces from the medial left foot or medial right foot are negative. The Y-axis reported anterior forces as positive values and posterior forces as negative value. The Z-axis recorded vertical forces. All force values were normalized for subject bodyweight in order account for body mass when comparing between subjects.

APPENDIX E

Biodex System II Isokinetic Dynamometer

The participant will be asked to lie supine or on his or her side on an isokinetic dynamometer. A standard knee attachment device was secured to the leg so the pad was placed between the knee and the hip. The research participant was be required to perform 3 isometric contractions (contracting the muscles without moving the joint) to assess the maximum strength for the hip flexion, hip extension, hip adduction, and hip abduction.

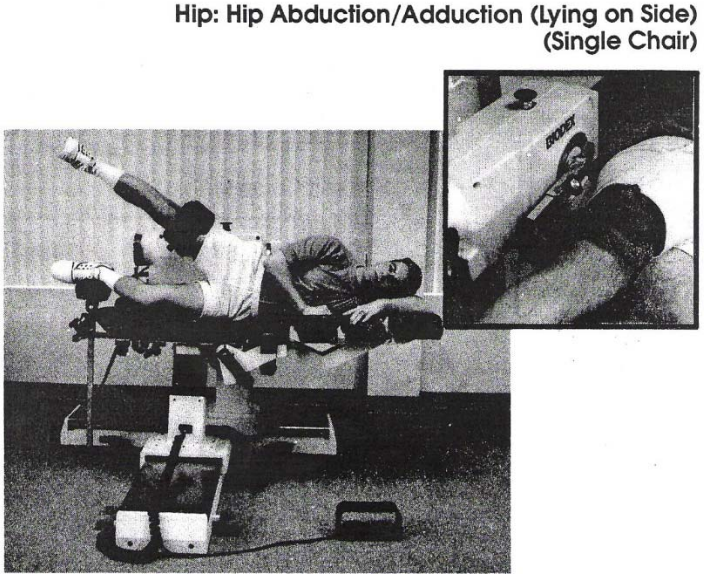


Figure 1. Hip Abduction/Adduction Strength Testing Protocol. Pictures provided by the Biodex handbook.

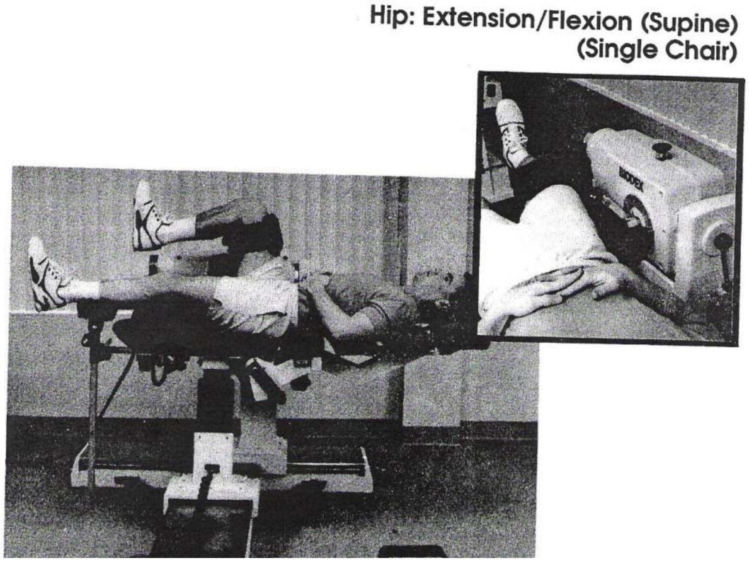


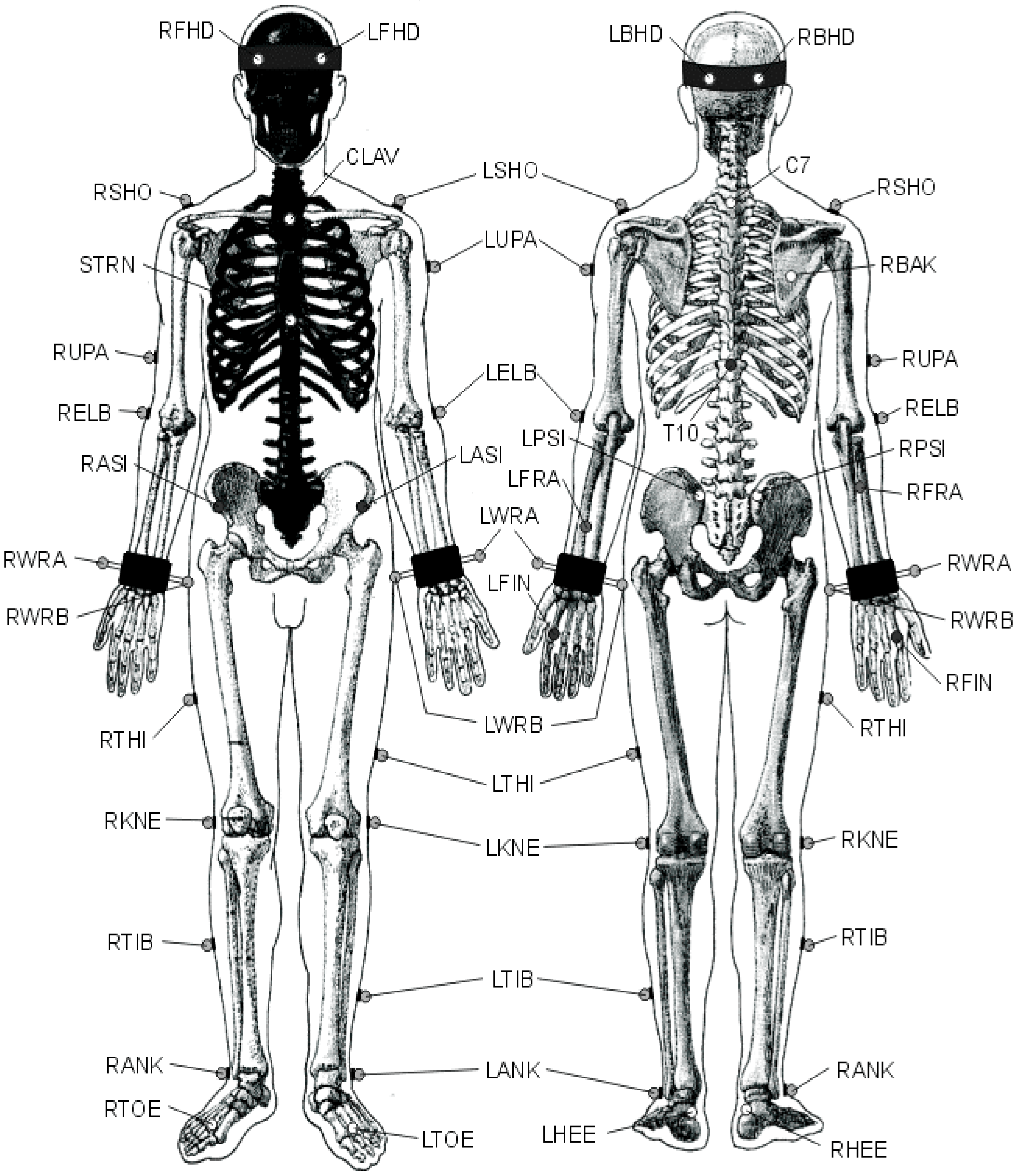
Figure 2. Hip Flexion/Extension Strength Testing Protocol. Pictures provided by the Biodex handbook.

APPENDIX F

Retroflective Marker Placement Based On VICON Plug-In Gait Model

Placement of the reflective markers on the participants based on the Plug-in-Gait Marker Placement Protocol for use with the VICON motion analysis system.

Plug-in-Gait Marker Placement



The following describes in detail where the Plug-in-Gait markers should be placed on the subject. Where left side markers only are listed, the positioning is identical for the right side.

Upper Body

Head Markers

LFHD	Left front head	Located approximately over the left temple
RFHD	Right front head	Located approximately over the right temple
LBHD	Left back head	Placed on the back of the head, roughly in a horizontal plane of the front head markers
RBHD	Right back head	Placed on the back of the head, roughly in a horizontal plane of the front head markers

The markers over the temples define the origin, and the scale of the head. The rear markers define its orientation. If they cannot be placed level with the front markers, and the head is level in the static trial, tick the "Head Level" check box under options on “Run static model” in the pipeline when processing the static trial. Many users buy a headband and permanently attach markers to it.

Torso Markers

C7	7th Cervical Vertebrae	Spinous process of the 7th cervical vertebrae
T10	10th Thoracic Vertebrae	Spinous Process of the 10th thoracic vertebrae
CLAV	Clavicle	Jugular Notch where the clavicles meet the sternum
STRN	Sternum	Xiphoid process of the Sternum
RBAK	Right Back	Placed in the middle of the right scapula. This marker has no symmetrical marker on the left side. This asymmetry helps the auto-labeling routine determine right from left on the subject.

C7, T10, CLAV, STRN define a plane hence their lateral positioning is most important.

Arm Markers

LSHO	Left shoulder marker	Placed on the Acromio-clavicular joint
LUPA	Left upper arm marker	Placed on the upper arm between the elbow and shoulder markers. Should be placed asymmetrically with RUPA
LELB	Left elbow	Placed on lateral epicondyle approximating elbow joint axis
LFRA	Left forearm marker	Placed on the lower arm between the wrist and elbow markers. Should be placed asymmetrically with RFRA
LWRA	Left wrist marker A	Left wrist bar thumb side
LWRB	Left wrist marker B	Left wrist bar pinkie side

The wrist markers are placed at the ends of a bar attached symmetrically with a wristband on the posterior of the wrist, as close to the wrist joint center as possible.

LFIN	Left fingers	Actually placed on the dorsum of the hand just below the head of the second metacarpal

Lower Body

Pelvis

LASI	Left ASIS	Placed directly over the left anterior superior iliac spine
RASI	Right ASIS	Placed directly over the right anterior superior iliac spine

The above markers may need to be placed medially to the ASIS to get the marker to the correct position due to the curvature of the abdomen. In some patients, especially those who are obese, the markers either can't be placed exactly anterior to the ASIS, or are invisible in this position to cameras. In these cases, move each marker laterally by an equal amount, along the ASIS-ASIS axis. The true inter-ASIS Distance must then be recorded and entered on the subject parameters form. These markers, together with the sacral marker or LPSI and RPSI markers, define the pelvic axes.

LPSI	Left PSIS	Placed directly over the left posterior superior iliac spine
RPSI	Right PSIS	Placed directly over the right posterior superior iliac spine

LPSI and RPSI markers are placed on the slight bony prominences that can be felt immediately below the dimples (sacro-iliac joints), at the point where the spine joins the pelvis.

SACR	Sacral wand marker	Placed on the skin mid-way between the posterior superior iliac spines (PSIS). An alternative to LPSI and RPSI.

SACR may be used as an alternative to the LPSI and RPSI markers to overcome the problem of losing visibility of the sacral marker (if this occurs), the standard marker kit contains a base plate and selection of short "sticks" or "wands" to allow the marker to be extended away from the body, if necessary. In this case it must be positioned to lie in the plane formed by the ASIS and PSIS points.

Leg Markers

LKNE	Left knee	Placed on the lateral epicondyle of the left knee

To locate the "precise" point for the knee marker placement, passively flex and extend the knee a little while watching the skin surface on the lateral aspect of the knee joint. Identify where knee joint axis passes through the lateral side of the knee by finding the lateral skin surface that comes closest to remaining fixed in the thigh. This landmark should also be the point about which the lower leg appears to rotate. Mark this point with a pen. With an adult patient standing, this pen mark should be about 1.5 cm above the joint line, mid-way between the front and back of the joint. Attach the marker at this point.

LTHI	Left thigh	Place the marker over the lower lateral 1/3 surface of the thigh, just below the swing of the hand, although the height is not critical.

The thigh markers are used to calculate the knee flexion axis location and orientation. Place the marker over the lower lateral 1/3 surface of the thigh, just below the swing of the hand, although the height is not critical. The antero-posterior placement of the marker is critical for correct alignment of the knee flexion axis. Try to keep the thigh marker off the belly of the muscle, but place the thigh marker at least two marker diameters proximal of the knee marker. Adjust the position of the marker so that it is aligned in the plane that contains the hip and knee joint centers and the knee flexion/extension axis. There is also another method that uses a mirror to align this marker, allowing the operator to better judge the positioning.

LANK	Left ankle	Placed on the lateral malleolus along an imaginary line that passes through the transmalleolar axis
LTIB	Left tibial wand marker	Similar to the thigh markers, these are placed over the lower 1/3 of the shank to determine the alignment of the ankle flexion axis

The tibial marker should lie in the plane that contains the knee and ankle joint centers and the ankle flexion/extension axis. In a normal subject the ankle joint axis, between the medial and lateral malleoli, is externally rotated by between 5 and 15 degrees with respect to the knee flexion axis. The placements of the shank markers should reflect this.

Foot Markers

LTOE	Left toe	Placed over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot
LHEE	Left heel	Placed on the calcaneous at the same height above the plantar surface of the foot as the toe marker

APPENDIX G

Comprehensive Examination Manuscript

THE EFFECTS OF GENDER ON THE MUSCLE ACTIVATION, KINETICS AND KINEMATICS AT THE HIP DURING A JUMP, LAND, AND UNANTICIPATED CUT TASK IN ADOLESCENT BASKETBALL ATHLETES

By

Mikaela Boham

A dissertation comprehensive examination to be submitted
in partial fulfillment
of the requirements for the degree of
Doctorate of Education: Curriculum, Instruction and Foundational Studies, Kinesiology
and Athletic Training Education Emphasis
Boise State University

February, 2009

The Effects of the Muscle Activation, Kinetics, and Kinematics at the Hip during a Jump, Land, and Unanticipated Cut Task in Adolescent Male and Female Basketball Athletes

Mikaela Boham

Committee: Ron Pfeiffer, Jonathan Brendefur, John McChesney, Lynda Ransdell, and Michelle Sabick

ABSTRACT

Context: Females are two to eight times more likely to sustain an ACL injury than their male counterparts participating in the exact same sport. The primary mechanisms of noncontact injury reported for ACL injury involves landing from a jump, unanticipated change of direction, and/or deceleration activities.

Objective: The purpose of this pilot study was to determine if we can collect accurate and representative data to evaluate the effects of gender on a jump, land, and unanticipated cut in adolescent male and female basketball athletes.

Design: Cohort study from local club teams.

Setting: University Laboratory.

Participants: Three healthy adolescent basketball athletes (females, n = 2;males, n = 1).

Interventions: Each participant was instructed to jump over a barrier, land with each foot on an in-ground forceplate, and cut in a specific direction. The participants were instructed to make a side step (right light indicates the participant should cut to the right leading with the right leg, and using the left leg as a plant leg). Each subject performed fifteen (15) randomized jump, land, and unanticipated cutting maneuvers.

Main outcome measures: The peak electromyography (EMG) and ground reaction force (GRF) [normalized with body weight] were analyzed during the landing for the three cutting directions. Kinematic variables include joint angles at landing.

Analysis: Multiple analyses of variances (ANOVAs) were used to compare the means from the variables collected during testing procedures. However, with only three subjects, it is difficult to draw conclusions in regards to the statistical analysis of the study.

Results: At this time, the researcher reserves the right to withhold judgment on the results of this study. There were not enough subjects to make statistical comparisons and therefore conclusions as to what those results would mean.

Clinical relevance: This study may advance our understanding of potential muscle activation strategies about the hip during sport specific activities. Additionally, this study could provide support for the screening of hip strength during the pre-participation physical and the education and creation of training protocols to enhance the strength of the musculature surrounding the hip.

Keywords: anterior cruciate ligament; kinematics; kinetics; knee

Previous research has attempted to determine the impact gender has on an athlete's ability to perform athletic maneuvers with some promising results. When evaluating injuries overall, male and female athletes are at similar risk for injury. However, when evaluating the risk of injury for the lower extremity, researchers have discovered a phenomenon at the knee. Female athletes are at an increased risk of injury for the ligament of the knee as they are 2 to 8 times more likely to injure their anterior cruciate ligament (ACL) than their male counterparts (Anderson et al., 2001; Decker et al., 2003; Huston & Wojtys, 1996; Hutchinson & Ireland, 1995; Junge & Dvorak, 2004; McLean et al., 2003; Moeller & Lamb, 1997; Piasecki et al., 2003; Pollard et al., 2004; Powell & Barber-Foss, 2000; Rozzi et al., 1999; Slauterbeck et al., 2002; Toth & Cordasco, 2001; Wojtys et al., 2003; Wojtys et al., 1994). In particular, females are far more likely to sustain a noncontact ACL injury during sports participation requiring large amounts of acceleration, deceleration, jumping, landing and changes of direction (Decker et al., 2003; McLean et al., 2003; Moeller & Lamb, 1997; Slauterbeck et al., 2002; Toth & Cordasco, 2001). Possible risk factors associated with noncontact ACL injuries include environmental, hormonal, anatomical and neuromuscular (Anderson et al., 2001; Moeller & Lamb, 1997). Other authors postulate the increased risk of injury is more dependent on sport specific activities rather than gender (Cowley et al., 2006).

Studies have suggested females perform athletic activity utilizing less trunk flexion (DiStefano et al., 2005; Decker et al., 2003; McLean et al., 2004; Salci et al., 2004; Yu et al., 2006), less hip flexion (Decker et al., 2003; DiStefano et al., 2005; Ford et al., 2005; Jackson et al., 2008; Kernozek, et al., 2005; Kulas et al., 2008; McLean et al., 2004a; McLean et al., 2004b; Salci et al., 2004; Wikstrom, 2004; Yu et al., 2006), greater hip adduction (Jackson et al., 2008; Jacobs et al., 2007, Pollard et al., 2004), greater hip internal rotation (Jackson et al., 2008; Pollard et al., 2004), greater knee abduction (Barber-Westin et al., 2005; Jackson et al., 2008; Lephart et al., 2004, Pollard et al., 2004), and less knee flexion angles (Decker et al., 2003; DiStefano et al., 2005; McLean et al., 2004; Salci et al., 2004; Sell et al., 2006; Wikstrom et al., 2004; Yu et al., 2006) than their male counterparts.

In addition, researchers have suggested females land with a different landing strategy than male athletes. The female participants appear to sustain greater vertical landing loads during athletic activity (Kulas et al., 2008; Lephart et al., 2004). Weakness, muscular imbalance, and reduced flexibility have been identified as predisposing factors for injury (Starkey & Johnson, 2006). According to Wikstrom et al. (2004) women tend to land with the knee and hip in more extended positions and thus subject themselves to higher ground reaction forces per body weight during the impact of landing. The forces sustained by the hip during running have been documented to increase up to five times the body's normal weight suggesting impact loads possibly contributing to injuries of both muscle and bone (Prentice, 2009). Large impact loads can cause the body to be forced to rely on ligamentous restraint to prevent anterior slippage of the tibia on the femur as the loads placed on the muscles surpasses their abilities to restrain movement.

Recently research has focused on the hip to determine the amount of muscle activation available based on gender. Russell and colleagues (2006) have suggested no differences are noted between the genders when examining the gluteus medius (the primary hip abductor); however, Hart and colleagues (2007) suggest gluteus medius activity is significantly higher in male athletes than in female athletes. The gluteus medius is the primary hip abductor muscle and has been reported to provide support to the pelvis and hip during midstance of the gait (Anderson et al. 2003). The gluteus medius muscle has also been postulated to control femoral internal rotation movement during activity. Without significant muscular strength in the gluteus medius, an athlete might not be able to functionally control hip adduction and internal rotation (Hart et al., 2007). Researchers have suggested the noncontact ACL injury occurs during a specific point in the range of motion identified as the “point of no return” or the “position of no return” (Blackburn & Padua, 2008; Hart et al., 2007; Ireland, 1999; Jacobs et al., 2007). The “point of no return” is described as movements consisting of hip adduction and internal rotation, knee valgus and external tibial rotation, and subtalar pronation (Blackburn & Padua, 2008; Ireland, 1999). The risk of ACL injury has been highly associated with both hip adduction and internal rotation moments thus implying decreased muscular activity to the “point of no return” at the hip could expose the knee to injury (Hart et al., 2007; Ireland, 1999).

Previous research has suggested different muscle activation patterns and muscular imbalances might predispose female athletes to risk of knee injury. In addition, literature suggests women display lesser knee, hip and trunk flexion during gait and landing tasks compared to males (Decker et al., 2003; DiStefano et al., 2005; McLean et al, 2004b; Salci et al., 2004; Yu et al., 2006). These studies have suggested sagittal plane coupling of the hip and knee could be determining factors for the risk of ACL injury (Decker et al., 2003; DiStefano et al., 2005; McLean et al, 2004b; Salci et al., 2004; Yu et al., 2006). The purpose of the pilot study was to determine the effectiveness of the research methodology while collecting dissertation data and to determine if the collection of given variables (EMG, Kinetics, and Kinematics) within the given paradigm can be measured with accuracy. The purpose of the dissertation study will be to determine the effects of gender on a jump, land, and unanticipated cut in adolescent male and female basketball athletes to examine if an increased risk of ACL injury potential exists.

METHODS

Participants

Four healthy, adolescent basketball athletes participated in this study; however, one of the male subjects data was not collected for the MVIC value and therefore is not included in this analysis (*Males* (n = 1): age = 159 mo. (13.25 yrs); height = 175.9 cm (69.25 in); weight = 63.4 kg (139.5 lbs); BMI = 20.45; vertical jump = 48.3 cm (19 in); years playing basketball = 8; years playing club = 4; *Females* (n = 2): age = 167 mo. (13.92 yrs); height = 164 cm (64 in); weight = 55 kg (122 lbs); BMI = 21; vertical jump = 40 cm (16 in); years playing basketball = 6; years playing club = 3).

Any participant was eliminated if they reported: (1) a history of previous knee injury and/or lower extremity surgery, (2) pain in lower extremity immediately prior to testing, (3) any injury to the lower extremity in last 6 months, or (4) any neurological disorder. In addition, the participants were required to refrain from undertaking any exercise within 24 hours of the test session in an effort to eliminate the confounding factor of fatigue. All participants were required to be on a club team (with a minimum of 2 years of experience playing basketball. The subjects were in good physical condition and accustomed to participating in running sprinting, change of direction and deceleration speed drills during basketball practice and games. All subjects signed an informed ascent document and parents signed an informed consent document, which was approved by the Boise State University Institutional Review Board (Approval # BM 103-09-002).

Subject Preparation

Arthropometric measurements (height, weight, age [date of birth], years of playing experience and dominant limb of the subjects) were obtained from the participant. The dominant limb was self reported by the participant by asking the participant which leg they would prefer to use to kick a soccer ball. The participants were asked to wear specific types of clothing to gather proper data. Participants were asked to wear tight fitting clothing, but were allowed to wear baggy shorts over spandex.

The skin superficial to the muscle belly of right rectus femoris, left rectus femoris, right biceps femoris, left biceps femoris, right adductor longus, left adductor longus, right gluteus medius, and left gluteus medius was prepared by cleaning it with isopropyl alcohol. Electrodes were oriented parallel to muscle fiber direction. Approximately 32 reflective markers were used to create a total body image for three-dimensional analysis of athletic maneuvers. The reflective markers were attached to research participant's body using double sided tape and/or elastic bands.

Instrumentation

A 6-channel surface Electromyography (EMG) system (BTS Free EMG) was used to collect electrical activity of the bilateral muscles surrounding the hip from small diameter (12 mm), round, silver/silver chloride, bipolar, preamplified electrodes (Myotronics, Inc., Kent, WA). The participants were fitted with an avalanche beacon harness to hold the wireless transmitter on the body while the subject performed dynamic jumping and running activities. All EMG placements are following the protocols listed in *Introduction to Surface Electromyography* written by Jeffry R. Cram & Glenn S. Kasman (Cram & Kasman, 1998). EMG signals were sampled at 1000 Hz and interpreted by MYOLAB and processed with MATLAB software (The MathWorks Inc.) EMG signals were automatically high pass filtered with a cut off at 15.9 Hz by the BTS Free EMG unit prior to the signal being sent to the patient unit and then the lab computer. To represent muscle activity during the jump, land, and unanticipated cutting maneuver a root mean square (RMS) algorithm used a 20-sample moving average.

A VICON (VICON Motion Systems, Lake Forest, CA) motion analysis system consisting of 6 infrared cameras controlled by Nexus software (VICON Motion Systems, Lake Forest, CA) provided joint position data during the jump, land and unanticipated cut maneuvers. Each camera was calibrated with motion analysis equipment and then again with a static trial for each participant. If calibration values fall above a desired range (0.2 pixel image error) then the calibration were performed again.

Two multiaxis inground forceplates (Kistler, Type 9821C) collected ground reaction forces. The force plates were sampled at 1250 Hz. Landing force was defined for each leg as the moment when the forceplate detected any vertical component (Fz greater than 20 N) of a ground reaction. The force plate was oriented so the X values represent medial and lateral components of landing with all values to the left (forces coming from the lateral left foot or medial right foot) being positive, and force values to the right (forces coming from the medial left foot or lateral right foot) being negative. The force plate's Y axis reports the anterior to posterior force moments with anterior forces being reported with positive values and posterior forces reported with a negative value. The Z-axis is the vertical ground reaction forces, which were normalized for subject bodyweight in order for comparison between subjects *(Appendix B)*.

The Biodex Isokinetic machine (Biodex System II) was used to assess Maximum Voluntary Isometric Contractions (MVIC) to determine the electromyography (EMG) analysis. The Biodex was used as a stationary force to collect EMG data from an isometric contraction for both the right and left leg in the four motions of the hip. The lever arm of the isokinetic dynomemeter was held stationary at 135° during isometric testing of the hip for flexion, extension, abduction and adduction.

Testing Procedures

Each participant reported to the BSU Center for Orthopaedic and Biomechanic Research lab to complete a one-time data collection. The minor participants read and signed an informed ascent to participation before data collection *(Appendix C)* and parental consent form *(Appendix D).* Each participant was instructed to jump over a barrier, land with each foot on an inground forceplate, and side cut in a specific direction. As the participant jumped over the barrier, a laser light triggered one of the three cutting directions to appear on a board while the participant was still in the air. Upon landing, the participants were instructed to make a side step (right light indicates the participant cut to the right leading with the right leg and using the left leg as a plant leg). The three directions were: 30° degrees to the right, straight ahead, or 30° degrees to the left. A total of fifteen (15) randomized jump, land, and unanticipated cut were performed by each participant.

Maximum Vertical Leap

The participant was asked to perform 3 maximum vertical jumps to calculate a vertical jump height. A Vertec vertical jump analysis was used to calculate the average maximum vertical leap for an individual participant. The average maximum vertical jump was used to determine the height of the hurdle, as the barrier was placed at 50% of the calculated average maximum vertical jump height. The hurdle was placed in front of the two ground reaction force plates as an obstacle the participant must avoid when performing the jump maneuver.

Jump, Land, and Unanticipated Cut Assessment

The participants were introduced to the jump, land and unanticipated cut maneuver. The athlete was given several practice attempts to master the skill of jumping over a barrier to land on a force plate. The participant was required to land with a single foot on each of the force plates. The force plate jump distance was approximately 120-150 cm for each participant and designated by a line on the ground. The force plates are embedded into the ground of the lab facility and therefore are immovable. The participant was shown a light to direct the cutting movement in one of three directions (30° to the right, straight ahead, 30° to the left). Upon landing, the participants were instructed to make a side step (right light indicates the participant should cut to the right leading with the right leg and using the left leg as a plant leg). The athletes were tested with fifteen jumps, 5 jumps in each of the directions presented in a randomized fashion. After the participants finished the athletic protocol, the electrodes and reflective markers were removed from the participant. The participant and his or her parent was debriefed and dismissed from the research study.

Isokinetic Assessment

The participants were fitted and tested on the Biodex II isokinetic machine to measure the maximum voluntary isometric contraction of the four muscles of the hip (gluteus medius, hamstrings, quadriceps, adductor longus). The research participant was then fitted on the Biodex II isokinetic machine to test hip strength. The participant was asked to lie on an isokinetic machine with a standard knee attachment device secured to the leg so the pad was placed between the knee and the hip. The research participant was required to perform 5 repetitions of an isometric contraction (contracting the muscles without moving the joint) to assess the maximum strength for the hip flexion, hip extension, hip adduction, and hip adduction.

DATA PROCESSING

Muscle Activation Processing

MYOLAB software was used to process raw EMG data after acquisition. MVIC mean amplitude muscle activity was calculated by determining the mean activity for each MVIC trail for each muscle. The mean across the MVIC trials was determined and used to normalize the muscle activity data collected during the side-step cutting task for each respective muscle. 100 to make the normalized mean amplitude muscle activation data during the land, and cut maneuver expressible as a % MVIC multiplied this normalized value.

Motion Analysis Kinematics Processing

Knee and hip kinematic data was analyzed using MATLAB. The point of ground contact will allow the maximum knee flexion, hip flexion, knee valgus/varus, hip valgus/varus angles were determined for each cut direction to be determined for statistical analysis.

Ground Reaction Force Processing

Peak GRF was calculated using MATLAB. Peak vertical GRF was defined as the maximum value of the VGRF (Vertical Ground Reaction Force). Body weight was used to normalize peak VGRF data. Peak sagittal (anterior/poster) and frontal (medial/lateral) was defined as the maximum value of each measure, for each the right and left leg as determined by the data collected from the force plate during ground contact.

STATISTICAL ANALYSIS

The independent variable was gender. The kinetic dependent variables was peak anterior/posterior ground reaction forces during landing; peak medial/lateral ground reaction forces during landing; and peak vertical ground reaction force during landing. The kinematic dependent variables were joint angles at the hip and knee, including flexion, extension, abduction or adduction. Dependent measures for muscle activation included normalized mean amplitude muscle activity (percentage of maximal voluntary isometric contraction [% MVIC]) for the right vastus medialis (RVM), left vastus medialis (LVM), right biceps femoris (RBF), left biceps femoris (LBF), right adductor longus (RAL), left adductor longus (LAL), right gluteus medius (RGM), and left gluteus medius (LGM) of both the right and left lower extremity.

Study Design

EMG

Statistical analyses were performed using a 3 [cut direction] x 2 [gender] factorial ANOVA. An a-priori α level of 0.05 was set for determining statistical significance. Dependent samples t-tests were performed to investigate significant main effects and interactions for each of the muscles used in the study (8 EMG leads).

Separate statistical analyses were performed using an independent samples t-test to determine the differences between the EMG signals for the left and right hips.

Kinematics

Multiple analyses of variances (ANOVAs) were used to determine the effect of joint kinematics on cut direction and gender. Post hoc t-tests were utilized to determine the differences among the significant variables. An a-priori α level of 0.05 was set for determining statistical significance.

Separate statistical analyses were performed using an independent samples t-test to determine the differences between the EMG signals for the left and right hips.

Kinetics

Multiple analyses of variances (ANOVAs) were used to determine the effect of joint kinematics on cut direction and gender. Post hoc t-tests were utilized to determine the differences among the significant variables. An a-priori α level of 0.05 was set for determining statistical significance.

Separate statistical analyses were performed using an independent samples t-test to determine the differences between the EMG signals for the left and right hips.

RESULTS, DISCUSSION, CONCLUSION

The intended purpose of this pilot study was to determine if the research methodology could be effectively implemented and to determine if the data collections for the variables (EMG, Kinetics, and Kinematics) can be collected accurately and with the intended purpose of being able to answer the research questions.

The researcher would like to withhold the results, discussion and conclusion analysis until the more data has been collected from future participants. A limited subject pool does not illustrate a representative sample of the desired population of study. The results produced from this small subject pool will be limited to the small population in which we studied. In addition, it is difficult to determine significance in a small sample size. With the completion of the rest of the dissertation, I feel confident we will have enough subjects to draw appropriate conclusions from the statistical measures.

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Subject Demographics

Subject	Gender	Age (In Mo. @ testing)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Year in School	Vertical Jump (cm)	Dominant Limb	Years Playing Basketball	Playing Club Basketball
1	Female	163	163.83	51.36	19.09	8th Grade	35.56	Right		
2	Male	159	175.90	63.41	20.45	7th Grade	48.26	Right	8	4
3	Male	161	165.10	50.00	18.30	7th Grade	40.64	Right	8	3
4	Female	170	163.20	59.09	22.14	8th Grade	44.45	Right	6	3
5	Male	202	185.42	79.55	23.09	11th Grade	59.69	Right	8	5
Average		171	171	61	21		46		8	4

Pilot Subject 1 EMG Data Template

	MVC Signal Maximum Average - Based on RMS							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	R Rectis
Average	**0.12**	**0.04**	**0.16**	**0.04**	**0.08**	**0.08**	**0.12**	**0.08**
Stdev	0.00	0.01	0.04	0.00	0.00	0.01	0.00	0.00

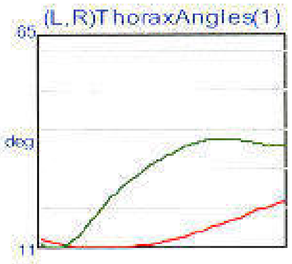
	Left Run Maximum Average - Based on RMS (Phase 1)							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	L Rectis
Average	0.23	0.04	0.17	0.14	0.17	0.05	0.06	0.17
Stdev	0.32	0.08	0.23	0.19	0.23	0.08	0.14	0.23
%MVC	**189.24**	**101.65**	**105.66**	**312.82**	**206.33**	**69.66**	**50.38**	**207.52**
	Left Run Maximum Average - Based on RMS (Phase 2)							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	L Rectis
Average	0.39	0.07	0.19	0.34	0.24	0.13	0.06	0.19
Stdev	0.54	0.16	0.26	0.54	0.33	0.18	0.14	0.26
%MVC	**321.03**	**199.58**	**121.59**	**758.56**	**298.31**	**166.14**	**51.61**	**238.81**

	Center Run Maximum Average - Based on RMS (Phase 1)							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	L Rectis
Average	0.09	0.07	0.13	0.05	0.06	0.08	0.06	0.13
Stdev	0.13	0.11	0.20	0.10	0.12	0.12	0.08	0.20
%MVC	**75.05**	**201.14**	**80.05**	**105.36**	**68.70**	**110.02**	**46.72**	**157.22**
	Center Run Maximum Average - Based on RMS (Phase 2)							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	L Rectis
Average	0.16	0.13	0.17	0.09	0.14	0.12	0.11	0.17
Stdev	0.23	0.17	0.24	0.21	0.26	0.16	0.15	0.24
%MVC	**131.27**	**343.35**	**110.31**	**208.65**	**174.52**	**157.51**	**87.60**	**216.66**

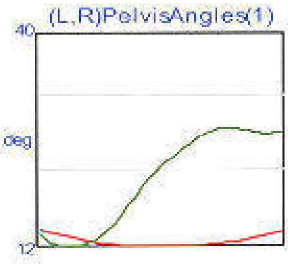
	Right Run Maximum Average - Based on RMS (Phase 1)							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	L Rectis
Average	0.13	0.07	0.16	0.14	0.18	0.13	0.10	0.16
Stdev	0.21	0.10	0.22	0.19	0.25	0.18	0.14	0.22
%MVC	**103.07**	**191.09**	**103.63**	**310.50**	**228.36**	**177.24**	**84.01**	**203.53**
	Right Run Maximum Average - Based on RMS (Phase 2)							
	L Biceps	L Adductor	L Glute	L Rectis	R Biceps	R Adductor	R Glute	L Rectis
Average	0.18	0.12	0.22	0.17	0.25	0.13	0.10	0.22
Stdev	0.18	0.12	0.22	0.17	0.25	0.13	0.10	0.22
%MVC	**147.79**	**316.63**	**139.84**	**391.15**	**311.93**	**177.24**	**86.72**	**274.65**

		Left Run Event Markers					Center Run Event Markers					Right Run Event Markers				
		Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5
Left Leg	Event 1	1848	2084	5	5	5	1784	1904	5	5	5	1648	1848	5	5	5
	Event 2	2096	2436	10000	10000	10000	1996	2108	10000	10000	10000	2024	2392	10000	10000	10000
Right Leg	Event 1	1840	2080	5	5	5	1780	1944	5	5	5	1632	1844	5	5	5
	Event 2	2316	2812	10000	10000	10000	2200	2392	10000	10000	10000	1832	2180	10000	10000	10000

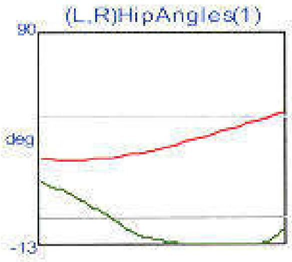
Kinematic Data Pilot Subject 2 (Male) – Trial 5 (Center Cut)



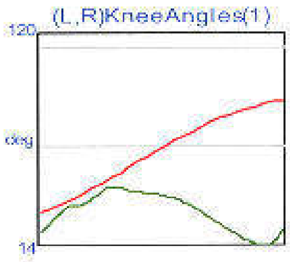
Thoracic
Angles
Sagittal Plane



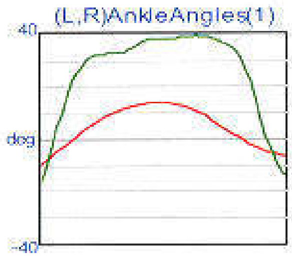
Pelvic Angles
Sagittal Plane



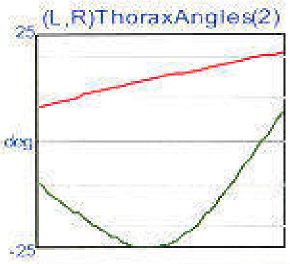
Hip Angles
Sagittal Plane



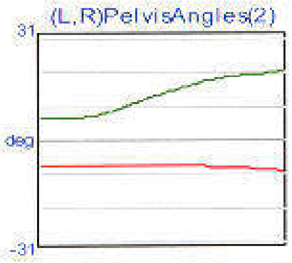
Knee Angles
Sagittal Plane



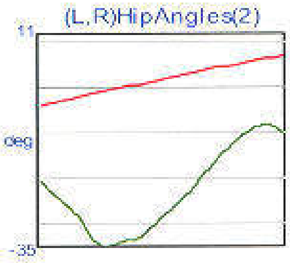
Ankle Angles
Sagittal Plane



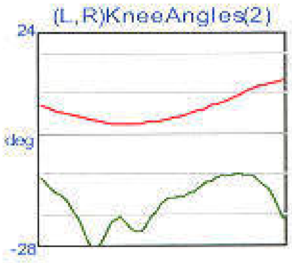
Thoracic
Angles
Frontal Plane



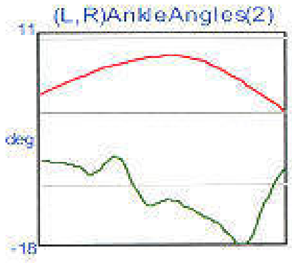
Pelvic Angles
Frontal Plane



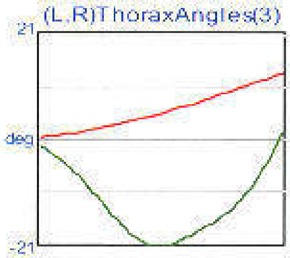
Hip Angles
Frontal Plane



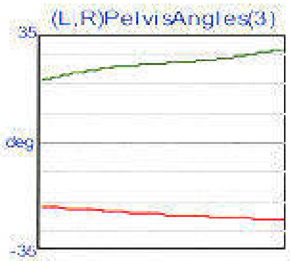
Knee Angles
Frontal Plane



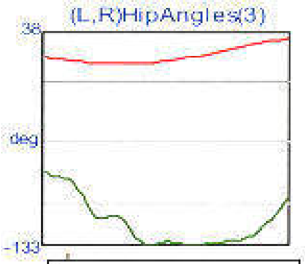
Ankle Angles
Frontal Plane



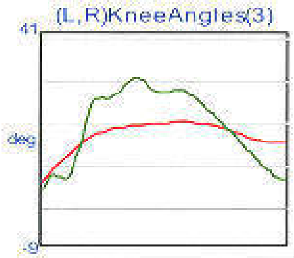
Thoracic
Angles
Transverse
Plane



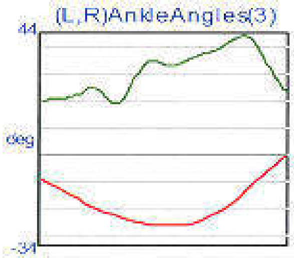
Pelvis Angles
Transverse



Hip Angles
Transverse



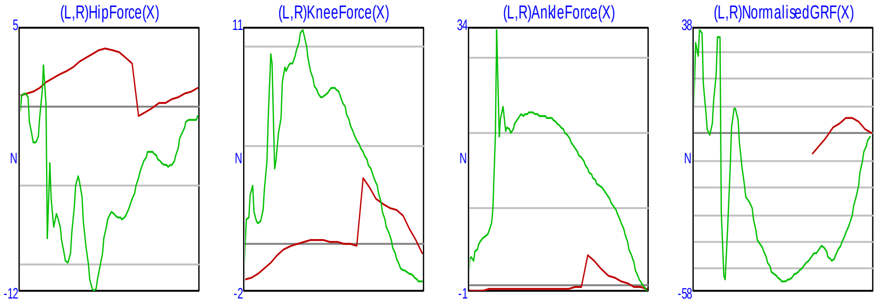
Knee Angles
Transverse



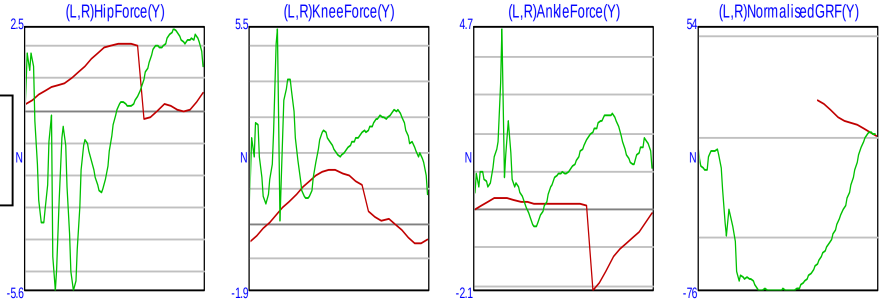
Ankle Angles
Transverse

Kinetic Data Pilot Subject 2 (Male) – Trial 5

Medial/Lateral GRF



Anterior/Posterior GRF



Vertical GRF