# The Effects of Exhaustive Exercise on ACL Injury Risk

by

Christopher Asplund

A THESIS

submitted to

Oregon State University

University Honors College

in partial fulfillment of
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degree of

Honors Baccalaureate of Science in Exercise and Sport Science
(Honors Associate)

Presented May 31, 2016
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Abstract approved:

_______________________________________________________

Samuel T. Johnson

**Background:** Certain knee movements place the anterior cruciate ligament (ACL) at risk of injury. These injuries commonly affect athletes especially at the end of events. The purpose of this study was to determine the effects of exhaustive exercise on frontal plane hip and knee mechanics during two tasks in order to better understand the risk factors for ACL injury. **Methods:** Forty, healthy, active female subjects performed drop jump landings (DJ) and side-step cut landings (SSC) before and after a 30-minute bout of exercise. The following frontal plane dependent variables were collected during the tasks: maximum hip and knee moments, angle at initial contact, maximum angle prior to maximum knee flexion, and overall displacement. For each dependent variable a 2 (task) x 2 (time) repeated measures ANOVA was performed. **Results:** There were no significant interactions or main effects for time, but there was a significant main effect for task for hip angle at initial contact ($p=0.006$), hip angle displacement ($p=0.032$), and hip abduction moment ($p<0.001$). **Conclusion:** The subjects did not have changes in frontal plane mechanics in response to the exercise.

The SSC had greater hip abduction at initial contact, abduction moments, and overall adduction than the DJ.

Key Words: ACL, fatigue, biomechanics, knee abduction

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# CHAPTER 1

## Introduction

The anterior cruciate ligament (ACL) is one of the primary stabilizing ligaments of the knee. Injury involving the ACL is very common in athletics, estimated at more than 100,000 annual injuries in the United States alone (Prodomos et al., 2007). Rupture of the ACL has many short- and long-term consequences for an athlete. The short-term effects include pain, four to six months of physical rehabilitation, and likely longer than a year removed from athletics (Ardern et al., 2011).

The long-term effects include a significantly higher risk for a secondary rupture. In fact, prior ACL injury is one of the strongest predictors of ACL injury risk in the future (Alentorn-Geli et al., 2009). Another manifestation of these long-term consequences is an increased risk of osteoarthritis. Li et al. (2011) discovered changes in the cartilaginous matrix as early as one year after a reconstructive surgery. Mather et al (2013) stated that an individual that sustains an ACL rupture and meniscal tear has up to a 48 percent chance of developing osteoarthritis in ten years. These cartilaginous changes inevitably lead to an early onset of osteoarthritis of the reconstructed knee and often result in a total knee arthroplasty. Nearly one third of patients undergoing total knee arthroplasty have a history of knee surgery (Brophy et al., 2014). Further, Brophy et al. (2014) showed that a history of knee surgery leads to total knee arthroplasty at an age that was on average 15 years younger than those who had never undergone knee surgery. Furthermore, the financial cost of ACL injuries is a large concern as each reconstructive surgery is priced at 17,000 dollars, which compounds with other costs related to surgery and recovery to create a net cost of 7.6 billion dollars annually (Hewett et al., 2005; Mather et al., 2013).

Of all ACL injuries, seventy percent of them are due to non-contact mechanisms, meaning that the ACL was not injured from a direct blow or collision (McNair et al., 1990; Boden et al., 2000). This lack of direct mechanism of injury is part of the reason that ACL injuries need to be further investigated and understood. It is unknown why one athlete can perform the same task, whether it is a jump landing or change of direction, over and over again without injury, then experience an ACL rupture performing this exact movement. Without considering a change in the external environment, there is little explanation as to what causes an ACL injury during a particular athletic movement. Proper knowledge of the exact mechanism of injury would lead to a treatment or prevention strategy that could potentially limit the incidence of ACL injury. Another factor of ACL injury that is largely unexplained is why females are between three and eight times more likely to sustain an injury than males (Prodomos et al., 2007; Hewett et al., 2005; Boden et al., 2010).

Many different factors have been hypothesized to put females at greater risk of non-contact ACL injuries including anatomical, hormonal, and neuromechanical differences between males and females. Anatomical factors like pelvic dimensions, Q-angle, and the shape of the intercondylar notch have been investigated (Griffin et al., 2000). Yet, as the ability to modify these factors is minimal, intervention would be difficult and likely not beneficial to the athletes without the possibility of severe side effects (Alentorn-Geli et al., 2009). Hormonal factors have been investigated as a potential explanation for higher incidence of injury in the female population. The female sex hormones, specifically estrogen and progesterone, have shown to affect the physical properties of the ACL (Liu et al., 1996). This effect has shown to weaken the tensile strength of the ACL and thus others have also reported greater propensity towards rupture earlier in the menstrual cycle as well as inhibited ligament strength in the presence of estrogen and progesterone (Alentorn-Geli et al., 2009). Moreover, a decline in motor skill functioning was noted during the premenstrual phase indicating that the large amount of estrogen in the prior luteal phase affected neuromechanical function (Posthuma et al., 1987). Even so, like anatomical factors, the hormonal factors appear difficult to modify and manipulate without resulting in adverse side effects. In order to produce a meaningful hypothesis, the risk factors involved must be easily modifiable by the athlete.

The factor potentially most easily modified is neuromechanical control, specifically factors of how one performs movements, including jumping and cutting, in a way that limits risk of injury. During the dynamic movements that are involved in athletics, certain forces and positions of the knee seem to put the ACL at greater risk. Markolf et al. (1990) first introduced the idea of varying positions placing different tensile loads on the ACL. Through cadaveric manipulation and buckle transducers, they were able to determine that both external knee abduction and adduction moments caused increased stress on the ACL. Specifically, knee abduction moments placed greater stress on the ACL regardless of the knee flexion angle, unlike the stresses placed on the ACL by a knee adduction moment which was greatest at 0 degrees of knee flexion and lesser at 45 degrees of knee flexion. Therefore, regardless of the angle of the knee, there is increased stress on the ACL when a knee abduction moment is applied. This makes it more dangerous as a knee adduction moment would be safer for athletes if they landed in greater knee flexion. A later study by Markolf et al. (1995) also noted the additive properties of multiplanar forces – specifically frontal and transverse plane moments – further increased the stresses placed on the ACL.

Hewett et al. (2005) supported the idea that certain movements place the ACL at risk for injury. In a prospective study, the authors tested the biomechanics of 205 females on a drop jump landing. They then followed the subjects through a one of their respective sporting seasons. They found that the athletes who went on to injure their ACLs had 6.4 times greater difference in knee abduction moments (KAM) between their injured and uninjured legs than for athletes who did not suffer an ACL injury during the season. Also, these subjects had 8.4 degrees more of knee abduction at initial contact, and 7.6 degrees at maximum knee abduction angle, than those who did not injure their ACL. Based on these studies it appears that the neuromechanical pattern resulting in greater KAM and greater knee abduction angles during dynamic movements could potentially place the knee in a more vulnerable position.

Fatigue, as a result of athletic competition, also has been shown to potentially affect athletes' neuromechanical control and knee motion during movement (Cortes et al., 2012). There is evidence that the majority of ACL injuries occur at the end of bouts of competition; indicating fatigue may be a factor in ACL injury (Hawkins et al., 2001). The presence of fatigue could be common in the case of ACL injury because of a possible degradation in neuromechanical control with increased athletic fatigue. Further, fatiguing protocols in the laboratory have demonstrated a reduction in the ability of muscular force output (Thomas et al., 2015) and an increase in knee abduction moments in females, but not in males (Chappell et al., 2005).

Varying exercise protocols have been used to produce the type of fatigue that the researchers believe to best represent athletics and the demands placed on the knee during competition. Inducing fatigue through an exercise protocol allows one to determine the exact effects experienced by an athlete during competition or training. One could hypothesize that fatigue plays a role in an athlete's decreased ability to control his or her body through less muscle activity and force generation. A fatiguing protocol, comprised of exhaustive exercise, then makes it possible to study how an athlete performs in a situation with, for example, decreased force output and neuromechanical control. As these factors decrease, there would be increased chances of placing greater stress on the ACL during dynamic movement patterns. With less ability to neuromechanically control movements, athletes may be more susceptible to forces and knee motion that produces a position that places greater strain on the ACL.

If a link between fatigue, KAM, and ACL injury could be established it could possibly lead to development of prevention programs to limit this risk factor and decrease ACL injury occurrence. Prevention of the initial anterior cruciate ligament injury could be important in improving joint health and long-term function for an athlete. The purpose of this study was to determine the effects of a bout of exhaustive exercise on the frontal plane knee mechanics during double-leg jump and single-leg cutting tasks.

# CHAPTER 2

Literature Review

## Introduction

Injuries to the anterior cruciate ligament (ACL) result in both short-term and long-term effects. The short-term effects include pain, loss of physical ability both in the athletic realm and with activities of daily living, and often carry large financial and emotional burdens with the effort needed for a successful rehabilitation. In fact, full recovery usually requires twelve months of physical rehabilitation, at which point the joint may still present the person with deficits (Ardern et al., 2011). On top of that, Ardern et al. (2011) reported that only a third of athletes have returned to their prior level of competitive sport after twelve months. These remaining deficits can often become long-term problems ranging from chronic pain and limited abilities, to increased risk of developing osteoarthritis and requiring total knee arthroplasty.

ACL injuries have been estimated to occur at a rate of more than 100,000 injuries annually in the United States (Prodomos et al., 2007). Each reconstructive surgery is priced at 17,000 dollars, which compounds with other surgery-related and rehabilitative costs to create a net cost of 7.6 billion dollars annually (Hewett et al., 2005; Mather et al., 2013). If the treatment is limited to only rehabilitation without surgical repair, the net cost increases further to 17.7 billion dollars annually (Mather et al, 2013). However, the financial aspect is not the only costs that are a result of an anterior cruciate ligament injury, nor is it likely that after recovery, the individual will be symptom free. In fact, once reconstructed, an athlete is more likely to endure a second rupture (Paterno et al., 2012). A prior study following nearly 1500 athletes saw over five percent suffer a contralateral anterior cruciate ligament injury within five years of reconstruction, and just over four percent reinjured the previously reconstructed ligament (Shelbourne, Gray, & Haro, 2009). This only leads to a longer recovery and a greater likelihood of developing long-term symptoms.

One of the common chronic symptoms that individuals experience after an ACL reconstruction is the development of osteoarthritis. Osteoarthritis is one of the most common degenerative joint diseases and often leads to swelling, pain, and motion dysfunction and over time, can lead to fractures and bone spurs around the joint. This will lead to further pain and reduced quality of life beyond the acute effects of the ACL injury. Li et al. (2011) noted changes in the cartilage matrix indicating progression towards osteoarthritis of a reconstructed knee as early as one year after surgery.

The long-term effects continue as risk of OA increases with concurrent injury to the meniscus, a structure that is often damaged along with the ACL (Meyer et al., 2008). Specifically, without meniscal involvement, Mather et al. (2013) reported that the individual had a 0 to 13 percent chance of developing osteoarthritis in 10 years, but the risk increases to a 21 to 48 percent if an injury involves a meniscus (Mather, 2013). Ferretti et al. (1991) concurred that preservation of menisci is the most important factor when preventing osteoarthritis after knee surgery, but stated that some osteoarthritic change is inevitable in the long-term regardless of meniscal involvement due to severe damage from either the injury or the reconstruction.

On top of the greater risk of a secondary rupture and the development of arthritic changes, Brophy et al. (2014) showed that people who had a history of ligament reconstruction, were recipients of total knee arthroplasty on average 15 years before individuals without history of knee surgery. A total knee arthroplasty would only mean another rehabilitation and the potential for further functional limitations, which can become a greater financial burden as they have a significant impact on work status, earnings, and overall quality of life (Mather et al., 2013). Overall, the long length of recovery is only the first effect of an ACL injury and other effects will often transcend the athletic realm and persist further on in life.

Given the consequences of ACL injury, it is important to determine factors contributing to this injury in order to determine ways to limit the occurrence in athletics. Yet, the specific mechanism of this injury, nor how to prevent it, is fully understood. By determining what contributes to making an athlete more susceptible to injury, possible solutions can be produced to limit these contributing risk factors. A reduction in risk factors likely leads to a reduction in injuries and an overall increase in safety for athletes.

## Injury Mechanisms

ACL injuries are typically classified as being the result of either a contact or non-contact mechanism. The former occurs as the result of a direct blow to the knee from an outside force, such as a collision with another person, which causes the ACL to be stressed beyond its limits. Because it occurs from direct contact it is typically assumed to be less preventable. The second type, noncontact injuries, are of interest as they have the fewest explanations as to why they occur. Noncontact injuries occur as a result of a scenario that involves an increase in internal forces of the knee, but not necessarily due to an external force on the knee. Instead, these internal forces are usually a result of rapid deceleration, often associated with a sudden jump, cut or twist (UCSF Department of Orthopaedic Surgery, 2016). Through injury questionnaires, it has been reported that 70 percent of anterior cruciate ligament injuries to be non-contact (McNair et al.,1990; Boden et al., 2000). The deceleration phase of a landing or cutting maneuver places a great amount of stress on the anterior cruciate ligament as it attempts to prevent undesired knee motion. In particular, sports that involve these stressful maneuvers, specifically basketball, soccer and volleyball, are the ones that see the largest proportions of athletes with ACL injuries (Boden et al., 2000).

Yet, without an external force explaining the abnormal movement causing the injury, it remains in question as to what can explain the difference between an injury-provoking movement and a healthy movement. Athletic movements become particularly more dangerous when the fatigue associated with athletics is involved. When one cannot sufficiently use muscles to stabilize during an unanticipated movement due to fatigue, stress is placed on the knee in various directions and non-contractile soft tissues such as ligaments must absorb the strain that is normally absorbed by the contractile forces of the muscles. It is at this point that structures such as the ACL would be loaded. If the load became too great, a rupture would result.

Further question is brought upon the mechanisms of injury, as females are more likely than males of the same athletic population to suffer an ACL injury. The discrepancy between sexes ranges from a three to eight times greater rate of non-contact ACL injury in females than males (Cortes et al 2012, Boden et al, 2010). Multiple reasons exist to potentially explain the sex differences in ACL injury.

**Risk Factors for ACL injury**

Three general types of factors have been cited to explain susceptibility towards anterior cruciate ligament injury: anatomical, hormonal, and neuromechanical differences both between the sexes and between those who have sustained injuries and those who have not. This section will briefly describe these risk factors.

Anatomical differences have been proposed as a potential risk factor for ACL injury. While anatomical variances are unlimited between individuals, the more common differences noted to have effects on ACL injury include joint laxity from surrounding structures, intercondylar notch width, Q-angle size, pelvic width, and physiologic size and strength of the athlete's ACL. In a study of 140 participants, being both athletes and non-athletes of both sexes, females were shown to have far more joint laxity (Huston and Wojtys, 1996), which would then reduce their ability to stabilize the joint. Further, Huston and Wojtys (1996) stated that functional stability is obtained from geometry of the joint surfaces, the menisci, the secondary ligamentous stabilizers, and most significantly from the surrounding musculature (e.g., the quadriceps, hamstrings, and gastrocnemius). One of the main findings of the study was that females took longer to produce peak force within these stabilizing muscles, and once peak force was reached it was less than males when normalized for body weight. Weaker musculature would not allow females to absorb forces from movements sufficiently, thus increasing the likelihood of having to absorb forces with a supporting structure such as the ACL, leading to increased strain, and potential rupture.

In relation to the anatomy of the femur itself, as referenced in a review paper by Griffin et al. (2000), nine different studies addressed differences between healthy and ACL-injured athletes in the anatomical structure of the femoral notch. These studies noted that intercondylar notches of those who had a history of ACL injury were on average smaller, a similar trend to what is seen between sexes as females generally have a smaller notch width. A smaller femoral notch could increase the risk of ACL impingement and constriction of particular movements that would in turn place the ACL in greater strained positions. However, it was also stated that there was too much variability present in measurements to definitively conclude the risk. (Griffin et al., 2000).

Further examining the differences in the lower extremity anatomy between the sexes, pelvis width has been hypothesized to contribute to increased ACL injury risk by a change in the Q-angle. The Q-angle is formed by a line from the anterior superior iliac spine to the center of the patella, and the center of the patella to the tibial tubercle (Alentorn-Geli et al., 2009). Thus a larger pelvic width would consequently result in larger Q-angles. Buchanan (2003) described how a higher measure of Q angle could then place the knee at higher risk of knee abduction. With the femur in a slightly adducted position due to a wide pelvis, the knee at rest would naturally have to be more abducted. Knee abduction in particular has been shown to significantly load the ACL, both while static and during dynamic movements.

Lastly, when referencing specific anatomical differences noted between the sexes, Chandrashekar et al. (2005) determined that on average, females have ACLs of decreased length, volume, mass, and cross-sectional area in comparison to males. This would cause one to believe that lower forces and loads would be tolerated by a female's ACL before injury occurred. Chandrashekar (206) reported that ACLs from females were found to have a lower tensile strength than a male's ACL and would thus take less tensile force to induce an injury or rupture. When combined with several of the factors above, it is clear that a weaker ACL only magnifies the other anatomical differences between sexes that could potentially lead to greater ACL injury risk.

The differences in hormonal risk factors of ACL injury stem from the idea that the female sex hormones, namely estrogen and progesterone, can cause changes in the composition and mechanical properties of the ACL (Liu et al., 1996), as well as causing effects in the neuromuscular system (Lebrun, 1994, Posthuma et al., 1987). The higher rate of ACL injury in female populations leads one to believe that there is a difference between the sexes, which could stem from the interactions of different hormonal levels. Specifically, Liu et al. (1996) determined that both fibroblast proliferation and the synthesis of collagen decrease as concentrations of estradiol, an estrogen sex hormone, increase. Naturally, females would have higher levels of estradiol and potentially less collagen and fibroblast activity. Estrogen sex hormone levels are highest towards the middle of the menstrual cycle, immediately before ovulation, and again during the luteal phase after ovulation. These points during the cycle also accounted for three quarters of ACL injuries noted by Moller-Nielson and

Hammar (1991). Through the cyclic stages, Lebrun (1994) saw differences in isokinetic strength and ability to exercise at aerobic and anaerobic capacities. Moreover, decreased motor skills assessed through in functional tasks during the premenstrual phase indicate that the large amount of estrogen in the prior luteal phase affected neuromuscular function (Posthuma et al., 1987).

Both anatomical and hormonal factors have potential to explain differences in ACL injury and help identify those at higher risk of injury. Yet, due to the fact that they cannot be easily modified, it is difficult to determine a solution that reduces risk and prevents injury in competition. Hewett et al. (2005) describes these factors as "nonmodifiable by nature" specifically because the extent of modification without adverse effects is unknown. It is very unlikely that either the anatomical or hormonal risk factors could be altered successfully in an athlete without causing other dramatic effects. This leaves the third factor mentioned by Hewett et al. (2005), neuromechanical factors, as the only ones that have potential for ACL injury contribution as well as the ability to be potentially modified through interventions to reduce risk.

**Neuromechanical Risk Factors**

The neuromuscular control aspect of movement, particularly as it relates to biomechanics, is referred to as neuromechanical factors. This could contribute a significant amount to the risk seen during the movements of jumping and cutting. This is especially of interest as the neuromechanics of an athlete are more likely able to be modified than the hormonal and anatomical risk factors of ACL injury.

A non-contact ACL injury likely has many factors that play in to the risk of the injury. During dynamic movements that are often a part of athletics, it is suggested that a particular position or motion of the knee place the ACL at a higher load and this greater stress causes a rupture (Chaudhari and Andriacchi, 2006). While in athletics, it is unlikely that a movement occurs in only one plane or has only one moment acting on a joint, frontal plane motion is commonly researched as a component of ACL injury mechanisms.

Markolf et al. (1990) first began studying the magnitudes of forces placed on the ACL in different positions using knees from cadavers. They placed buckle transducers in series with the ACL in order to measure the force applied to the ligament. A mechanical apparatus was designed to apply constant loads in different planes on the knee through attachments on the tibia and femur. The transducers were then able to quantify the stress on the ACL at times of various applied loads in multiple planes. Their study revealed that frontal plane kinetics, i.e., knee adduction and abduction moments, produced an increase in tension on the ACL. Interestingly, they reported that strain on the ACL caused by a knee adduction moment was less with greater knee flexion. In contrast, the ACL sustained elevated strain with a knee abduction moment from zero to 45 degrees of flexion. This increased force on the ACL during zero to 45 degrees of flexion could then be increased further with other coupled moments showing an additive effect on ACL strain (Markolf, et al., 1995). While a knee abduction moment alone can cause increased ACL strain, additional strain could be placed on the ligament with further anterior tibial force applied. Markolf et al. (1995) saw an additive effect of ACL strain with internal rotation torque added on to a previously applied anterior tibial force. They suspected that anatomical design allows collateral ligaments to resist fontal plane motion such as knee abduction moments. Then as an anterior tibial force is applied, the translational movement is resisted by the ACL. Therefore, with multiple planes involved, stress would be placed across more structures and increase the risk of rupture. Specifically, at any point with greater than ten degrees of flexion, Markolf et al. (1995) showed that a knee abduction moment would cause increased force on the ACL when applied along with an anterior tibial force. While flexion can alter the amount of force placed on the ACL, it was the interaction with the frontal plane movement that raised concern when ACL stress maintained at a high level when a valgus force was applied.

Shin, Chaudhari, and Andriacchi (2011) performed a similar study to that of Markolf, et al. (1995) in that they used a three-dimensional cadaveric knee apparatus to test the stress applied on the ACL from various motions and angles. The study incorporated loads on the knee to simulate single-leg standing while other moments were applied through the apparatus. Specifically, the study found that a combination of knee abduction moment and tibial internal rotation moment caused the greatest strain on the ACL. Even as the forces generated in the laboratory setting are likely lower than that experienced in an athletic setting, they were still able to generate ACL strain that was above the known limit for potential ACL rupture. They hypothesized that if the knee had been subject to the magnitude of forces produced during a full-speed athletic event, the strain placed on the ACL would be even higher above the threshold of rupture, increasing the likelihood of injury. The study did not show significant enough strain to produce injury for either knee abduction or tibial internal rotation alone.

Hewett, et al. (2005) supported the idea of a frontal plane factor after their study followed 205 female athletes over a total of 13 months, which was one season of their respective sports, and tracked ACL injuries. To analyze the subjects' biomechanics and knee motion during athletic tasks, a system of cameras and reflective markers were used. Prior to each athlete's season, 3D biomechanical data was collected while the athlete performed a drop off of a 31-centimeter tall box followed immediately by a maximal vertical jump. Ground reaction forces were collected through force plates during the initial landing off of the box. After data collection, the athletes were all monitored as they practiced and competed throughout their respective seasons. Of the 205 participants, nine sustained ACL injuries, all of which were non-contact in nature. The previously collected data were then compared between those who were injured during the season and those who were not. Females that injured their ACL displayed 8.4 degrees greater knee abduction angles at initial contact and 7.6 degrees greater of peak knee abduction angles than uninjured females. Knee abduction moments (KAM) were also 6.4 times greater when comparing differences between the two legs for those who sustained an injury. The KAM and knee abduction angles showed correlations with each other during the trials. A 20 percent greater ground reaction force measured in the athletes that would sustain an ACL injury was also noted. During the tested jump landings of the study, it appears as though an increased KAM and knee abduction angles can help to identify differences between athletes who will go to suffer an ACL injury.

While Hewett, et al. (2005) found significant increases in KAM for athletes who would go on to sustain an ACL injury, Kristianslund and Krosshaug (2013) suggest that the drop vertical jump task used in the study was not sufficient to produce significant KAM. The authors performed a comparison between drop vertical jumps and a sport-specific sidestep cut with 120 players from the Norwegian elite female handball series. The drop vertical jump task was nearly identical to that of Hewett, et al. (2005) except that it was off of a 30-centimeter box. The side-step cutting task was performed at an average of 3.4 meters per second and at an angle of 69 degrees. While a peak was observed in knee abduction moments just after initial contact during a sidestep cut, there was no peak at this point when performing a drop jump. In fact, the knee abduction moments during the cutting tasks were six times greater than those demonstrated during the drop jump task. A 5.8 degree greater knee abduction angle was also viewed in sidestep cuts in comparison to a drop jump. This suggests that an athlete with a measurable knee abduction moment during a drop jump task could have even higher moments when performing a cutting task. While a more sport-specific cutting task, as demonstrated in this study, may be more likely to produce knee abduction moments and angles that could lead to ACL injury than a drop jump, the drop task could produce greater motion in the sagittal plane which also has potential to stress the ACL. For this reason, both tasks lead to valid tests of movements that can increase ACL injury risk.

Since laboratory and cadaveric studies cannot generate the exact conditions experienced during an athletic event, they cannot be guaranteed to produce a perfect correlate to ACL rupture circumstances. Yet, other observations following ACL injury have led to further belief of the frontal plane injury mechanism. Quatman et al. (2011) noticed lateral tibial and femoral bruises and articular cartilage damage in a developed knee model used to produce representations of ACL injury. They hypothesized that this bruising was likely a product of lateral compression during knee abduction. The patterns determined by the study aligned with data in which Graf et al. (1993) described 71 percent of subjects showing a bone bruise after withstanding an ACL injury. This lesion that Graf et al. (1993) referred to consistently appeared on the lateral tibial plateau and lateral femoral epicondyle. The location suggests that a significant force brought together the lateral side of the knee joint, likely a knee abduction moment that then opened up the medial side of the joint surfaces.

If a study could combine these neuromechanical risk factors to point towards a particular movement, the knowledge could be used to develop programs to eliminate or reduce this risk factor. Hewett et al. (2005) hypothesizes that injury may be due to the fact that the event causing ACL injury may be too fast for reflexive contraction and muscle protection. Therefore, athletes would need to change the way that they land in order to protect themselves from motion that can result from an abducted knee positioning in order to avoid this risk factor. For example, a decrease in knee abduction moments through corrected jumping movements, as suggested by Hewett et al. (1996), could eliminate decrease the risk factors involved with ACL injury. Through the implementation of training to improve knee control throughout cutting and landing activities, Myklebust et al. (2003) showed that over three seasons women's ACL injury incidence could be reduced. This shows that through movement corrections and implemented training programs, ACL injury incidence could be reduced, while also limiting potentially high-risk movements (Hewett et al., 1996, Mykelbust et al., 2003). The possibility of "preprogramming" the athlete to land in a safer knee position after an unexpected load could reduce injury risk. This altering of mechanics relies on the fact that neuromechanical risk factors can be developed and fine-tuned through somatosensory and proprioceptive input used to map out efferent motor commands. With proper "preprogramming" to avoid identified harmful movements, we could dramatically decrease the incidence of ACL injury.

**Effects of Fatigue**

While identification of a particular movement pattern that places the ACL at greater injury risk is a step in the right direction, it also needs to be determined what causes these harmful movement patterns. There is evidence that a difference in an athlete's task performance exists between the start and end of an athletic event. Hawkins et al. (2001) established that a majority of non-contact injuries in soccer occur during the final 15 minutes of either half as well as an increase in injury occurrence with a longer athletic event duration. From this, it could be extrapolated that fatigue may be a factor in non-contact injury risk as an athletic event progresses and athletes fatigue. The presence of fatigue could be common in the case of ACL injury because of a possible degradation in biomechanics and neuromuscular control with increased athletic fatigue. Greater amounts of fatigue at this point in the event have been shown to reduce voluntary muscle contraction, impair muscle coordination, and limit the athlete's ability to stabilize the joint through dynamic movements

(Cortes et al., 2012). As several studies have looked at the effects of varying types of fatigue on lower extremity biomechanics, this section will compare the methods and results of fatigue's effects on neuromechanical risk factors.

Thomas et al. (2015) compared both healthy and ACL-reconstructed (ACLR) individuals in a study relating quadriceps strength and knee biomechanics before and after a fatiguing protocol. Quadriceps strength was measured through a single knee extension maximum voluntary isometric contraction while in 85 degrees of hip flexion and 90 degrees of knee flexion on an isometric dynamometer. Trials were repeated with rest until improvements were no longer made to ensure that the subject was accustomed to the equipment. Biomechanical data through reflective markers were collected three-dimensionally while the subject jumped with two feet over a 17-centimeter box and landed on one foot on a force plate, then they would immediately hop laterally off of their landing leg. During the exercise component, each subject would complete eight double-leg squats to at least 90 degrees of knee flexion and then perform three lateral hop landings. The lateral hop landings were recorded for biomechanical analysis. Fatigue was determined to be maximal when the subject could no longer perform five consecutive double-leg squats to 90 degrees of knee flexion without assistance. As a result of the fatigue protocol, it was noticed that both the healthy and ACLR groups had reductions in quadriceps strength when measured after fatigue had been determined. As fatigue increases throughout an athletic event, it is expected to see a decreased ability for the muscles to generate force output. Interestingly, the fatiguing protocol resulted in decreased isometric force production for both groups, but only the control group showed alterations in both knee flexion and abduction in response to the fatigue protocol. Participants in both groups showed less knee abduction moments and angles after fatigue, but more significantly in the control group. However, this finding was attributed to a declining exertion with fatigue meaning that the subjects did not push off as hard during the landing. The fact that the fatiguing protocol of squats being only in the sagittal plane was also a concern. Increased knee abduction has been attributed to fatigue of hip stabilizers in planes other than the sagittal plane. Therefore, as this protocol did not have a change-of-direction component, it could not have targeted the muscles that allow for increased knee abduction moments and angles with fatigue.

Chappell et al. (2005) completed a study of knee kinetics and kinematics through stop jump tasks both before and after exercise. The fatiguing protocol was comprised of five consecutive vertical jumps followed by an immediate 30-meter sprint and then repeated until volitional exhaustion. Five consecutive vertical jumps were also performed in between each post-fatigue jump trial to maintain a fatigued state. This resulted in a 96 percent increase in knee abduction moment of female subjects. It was noted that male subjects also saw a decrease of 43 percent in knee adduction moment post-fatigue. Thus, the female subjects in the study were at much higher risk of ACL injury than the males when landing in a flexed position. This was hypothesized by Chappell et al. (2005) to be a protective effect in the males to maintain a neutral knee position with increased fatigue, but it is not truly understood. If this is true, it can be expected that females lack this response, or do not utilize it to the same degree as male subjects. This produces the larger knee abduction moments with fatigue and a resultant greater stress on the ACL.

Chang et al. (2013) used another variation of repeated bouts of exercise to produce a sport-like fatigue. Each exercise bout consisted of five, six-minute cycles, to total 30 minutes of exercise with measurements of 3D biomechanical data before and after the entire exercise protocol. The cycles were broken up into 5 minutes of treadmill walking at an increasing incline of 0 percent to 15 percent and then 30 seconds of double-leg tuck jumps and 30 seconds of double-leg side-to-side leaps. No rest was given between cycles. It was performed two consecutive days, and was determined to induce a 21 percent decrease in quadriceps strength upon initial testing day for healthy individuals. The second day of testing had a smaller quadriceps strength reduction after fatigue (15 percent) indicating the possibility of an adaptive mechanism or that the full recovery had not been made from the previous day's exercise. A similar adaptive mechanism was proposed for subjects in the same study with reconstructed ligaments. The reconstructed-ACL subjects did not exhibit a decline in knee extension torque after the exercise bouts. While completing the post-fatigue test, jumping exercise was not required to maintain a fatigued state as it was by Chappell et al. (2005). This could potentially lead to less acute effects of fatigue being measured as recovery is likely to occur between trials after the exercise has stopped and contribute to some variance between the studies' results. However, the significant decline of knee extension torque after exercise in healthy subjects validates the bout as an exhaustive task. It could be used to produce fatigue in healthy subjects. With the decline in quadriceps strength, it can be expected that we would likely see modifications in landing biomechanics to account for the lost strength.

The use of a treadmill to elicit fatigue could be brought in to question as not producing specific enough fatigue to the dynamic fatigue resulting from an athletic event. Raja Azidin et al. (2015) addressed the concern of treadmill and sport-specific fatigue correlations, specifically in regards to soccer, by relating the two types of fatigue through biomechanical assessments. One was a treadmill exercise bout that incorporated 45 minutes of running on a treadmill while undergoing gradual changes in speed to mimic some acceleration and deceleration forces. The sport-specific exercise model included shuttle running, navigating through vertical poles, and varying velocities between walking, jogging, striding, and sprinting. Both exercise bouts had three-dimensional analysis of a 45 degree side-cutting maneuver recorded prior to exercise, at the conclusion of exercise, and 15 minutes after the conclusion of exercise. Raja Azidin et al. noted that while the sport-specific exercise simulation resulted in significantly higher heart rate increases and a higher rate of perceived exertion of the subjects, differences were not significant for peak knee abduction moments. This shows that treadmill induced fatigue can be used to accurately model changes in biomechanics due to fatigue in the lower extremities. While the subject may not feel the same systemic fatigue while on the treadmill as in an exercise bout more closely tied with a sport, the fatigue that alters mechanics is not significantly different for the two exercise types. This study did not push the subjects to volitional fatigue, but determined that these styles of exercise were more consistent with athletic scenarios as sports will not always push an athlete to volitional exhaustion.

Similar to Raja Azidin et al. (2015), Cortes et al. (2014) incorporated a definition of fatigue that was not volitional exhaustion. Fatigue in this study was defined as being the failure for the subject to complete three consecutive vertical jumps at 90 percent of the previously determined maximum vertical jump. It was also determined to be reached if the subject's heart rate rose to within 90 percent of an estimated heart rate maximum. This fatiguing protocol focused less on the aerobic components during the exercise bouts. The exercise consisted of three vertical jumps maintaining 90 percent of maximum vertical height, three parallel squats to 90 degrees of knee flexion, an agility drill running 5-10-5 yards, and 20 seconds of step ups on a 30-centimeter box at 200 beats per minute. Data were then collected after each exercise set and reported to be 100 percent fatigued after being unable to maintain the 90 percent of maximum vertical jump. The decrease of only ten percent of maximum vertical jump height can be related more to the expected athletic fatigue that Raja Azidin et al. (2015) discussed. When continuing to volitional fatigue, these studies expect the subject to display differently than one would in an athletic competition that does not push the athlete to this dramatic of an exertion level.

As was reported in several of the studies, regardless of the defining nature of the fatigue, or the specific protocol used to induce fatigue, an increase in fatigue often led to a change in the biomechanical patterns used by the subjects while jumping or landing. As Chang et al. (2014) showed significant effects in muscle strength, it is clear that a lasting effect was imparted as a result of the exercise. The lateral leaping and tuck jumps can also be expected to fatigue muscles other than the quadriceps that would likely be used to stabilize the knee during landing tasks. With the high incidence of injury during the moments in athletic events when fatigue is most common, it would be ideal to see exactly how fatigue changes an athlete's ability to control the neuromechanical risk factors associated with ACL injury. One would expect that fatigue would increase, neuromechanical risk factors would follow in suit, and the athlete would have a higher chance of ACL injury. If this process was confirmed, intervention protocols could target ways to limit this fatigue and engineer ways to prevent an athlete from adopting neuromechanical risk factors during athletic events.

# CHAPTER 3

## Methods

### Subjects

Forty healthy and physically active female volunteers between the age of 18 and 30 were recruited through fliers and announcements in university classes, as well as the Corvallis community. In order to be included in the study, each subject must have reported participating in an average of 150 minutes of moderate to vigorous activity each week, including participation in a jumping or cutting sport in the last six months. A subject was deemed ineligible for the study if she had any low back or lower extremity injury in the past six months hindering performance or had any previous surgery to her low back, hip, knee, or ankle.

### Experimental Procedures

Upon arrival at the Biomechanics Laboratory, the subject was instructed to change into spandex shorts, a tank top, and personal athletic shoes. She was informed of the procedures and risks of the study and then was asked to read and sign an informed consent form. The subject then provided information on her current and recent physical activity levels. Height and weight were measured in the laboratory using a stadiometer and scale, respectively. Prior to testing, each subject performed a five-minute warm-up on a stationary bicycle at her chosen pace. The subject's leg dominance was determined through a series of three tests: kicking a ball, stepping up a step, and recovering balance after a shove from behind (Hoffman et al., 1998). The dominant leg was defined as the one used for two of the three tests.

The subject was then instructed on the two tasks that they would be performing for the study. The first task was a drop jump. Subjects were instructed to jump off of a 30-centimeter box with both feet at the same time onto two side-by-side Bertec (Bertec Corporation, Columbus, OH, USA) force plates, with one foot landing on each plate and then immediately transition into a maximum vertical jump. The 30-centimeter box was placed 50 percent of the subject's height away from the edge of the force plates. (See Figure 1.)

The second task was a side-step cut as described by Frank et al (2011). The subject was positioned at a distance 50 percent of her height away from the force plates from even ground. She was instructed to jump over a 17-centimeter high hurdle that was placed half of the distance between the subject and the force plate (i.e., 25 percent of the subject's height). The subject took off with two feet to jump over the hurdle, landed on her dominant leg on the middle of one force plate, cut at 60 degree angle (based on arrows drawn on the floor), and sprint for several steps. (See Figure 2.)

Once oriented with the landing tasks, 27 reflective markers were placed on the subject's body. The markers were located bilaterally on the acromion process, anterior superior iliac spine, posterior superior iliac spine, first sacral vertebrae, greater trochanter, distal end of contracted rectus femoris, medial epicondyle of the femur, lateral epicondyle of the femur, anterior tibia, medial malleolus, lateral malleolus, distal end of the first metatarsal, distal end of the fifth metatarsal, and calcaneus. Markers were placed over shoes and spandex shorts when necessary.

First, a static trial was collected to calibrate the motion capture system with the anatomical measurements of the subject. Following completion of the static trials, the medial malleoli and medial epicondyle reflective markers were removed and the subject began collection of the kinematic data. Subjects were reminded of instructions for the tasks and then performed the drop jump task and the side-step cutting task. For both tasks, trials were performed with 30 seconds rest between trials until five acceptable trials were completed. For the drop jump landing, trials were deemed unacceptable if the subject did not land with her foot completely on the force plate, if her foot slid while landing and transitioning to the vertical jump, or if the vertical jump was not performed after the drop jump due to loss of balance or the subject forgetting to jump after landing. For the side-step cut, trials were unacceptable if the subject did not land completely on the force plate, did not take off simultaneously with both feet and land with one, or did not follow the correct 45-degree path in a sprinting manner.

Following the jumping tasks, the reflective markers were removed from the subject's body and the subject began the 30-minute exercise protocol used by Chang et al. (2014). The protocol consisted of five sets of five minutes of incline treadmill walking and one minute of jumping. The subject began walking on a treadmill at an individually chosen speed between 3.0 and 3.5 miles per hour at zero percent grade. At the end of each minute of walking, the treadmill would increase by one percent grade. The treadmill incline would continue to increase at a one percent grade up to a 15 percent grade as shown in Table 1. Each walking cycle lasted five minutes followed by 30 seconds of tuck jumps and 30 seconds of lateral leaps over a 17-

centimeter hurdle. The tucks jumps were completed by the subject bringing both knees up to her chest and then landing with two feet. For the lateral leaps, tape was placed on either side of the hurdle so the subject leaped a total distance of 75 percent of her height. The subject then began by jumping off one leg, over the hurdle and landing on the opposite leg. This was then repeated in the opposite direction. The subject was instructed to land as close to the tape as possible. Verbal encouragement was continually given during both of the jumping exercises. After both jumping sessions were completed, the subject returned to the treadmill to begin another 5-minute cycle of walking.

Heart rate and rating of perceived exertion (RPE) were collected at the beginning of each set. A heart rate monitor was fixed on a strap just below the subject's sternum. RPE was collected verbally with guidance from a chart representing the numbers. After completion of the fifth set of walking and jumping, the reflective markers were placed back on the subject as described previously. The subject then completed five acceptable trials of the drop jump landing and the side-step cut landing.

**Data Sampling and Reduction**

During the landing trials, a nine-camera Vicon Motus (Vicon, Incorporated, Centennial, CO, USA) motion capture system was used to collect kinematic data at 120 Hz. The force plate data were sampled at 1560 Hz. A Motion Monitor data acquisition system (Innovative Sports Training, Chicago, IL, USA) was used for the biomechanical model generation following import of the kinematic and kinetic data.

Joint centers were defined as the following: ankle – midpoint of the medial and lateral malleoli; knee – midpoint of the medial and lateral femoral epicondyles; and the hip – external landmarks as described by Bell et al. (1989). The local coordinate was defined with the positive x-axis directed anteriorly, the positive y-axis directed to the left, and the positive z-axis directed superiorly for the shank, thigh, and pelvis. Both the kinematic and force plate data were digitally low-pass filtered at 10 Hz using a 4$^{th}$ order zero-phase lag Butterworth filter with kinematic data time synchronized to the kinetic data and resampled at 1560 Hz. Euler angles, in a Y (flexion/extension), X' (adduction/abduction), and Z" (internal/external rotation) rotation sequence, were used to calculate joint angular positions using a right hand convention. Motion of the knee was defined as the shank relative to the thigh, and the hip as the thigh relative to the sacrum. Net internal joint moments of force at the knee and hip were calculated by combining the kinematic data, anthropometric data, and force plate data using an inverse dynamics solution within the Motion Monitor software (Gagnon & Gagnon, 1992).

A custom LabView program (National Instruments, Austin, TX, USA) was used to identify the following dependent variables of the study: hip abduction moment (mean of the first 100 ms), knee adduction moment (mean of the first 100 ms), and hip and knee angle at initial contact and maximum (peak) prior to peak knee flexion. Additionally, hip and knee displacement was calculated post-processing by subtracting the initial contact angle from the peak angle for the both joints. The mean values for each dependent variable were calculated for both tasks (i.e., side-step cut and drop jump) at each time (i.e., pre-exercise and post-exercise) for each subject. In some cases not all five trials were able to be used so the mean of the remaining trials was calculated.

**Data Analysis**

Of the 40 volunteers, 39 of them completed the study. Due to data collection errors, six more subjects were removed from the final analysis for a total of seven subjects removed prior to data analysis. One subject was removed due to her inability to complete the exercise protocol. Specifically, she needed to decrease the treadmill incline below ten percent in order to complete the 30 minutes of exercise protocol. Previous studies reported a significant change in muscular fatigue and energy expenditure when walking on a grade that is 10 percent or steeper (Chumanov et al., 2008; Slider et al., 2012; Wall-Scheffler et al., 2010). The other six subjects were removed due to errors during the data collection. The 33 subjects used for the study had an average age of 21.1 ± 1.7 years, an average height of 1.68 ± 0.08 meters, and an average body weight of 64.55 ± 7.94 kilograms.

Eight separate 2 (task) x 2 (time) repeated measure ANOVAs were completed for each dependent variable (shown in Table 2) using RStudio (0.99.896, Boston, MA, USA). The alpha level was set *a priori* at 0.05.

Table 1. Structure of the exercise protocol

| Set | Total Time | Treadmill Time and Incline | Tuck Jumps | Lateral Leaps |
|---|---|---|---|---|
| 1 | 0:00 – 6:00 | 5 minutes 0% - 4% | 30 seconds | 30 seconds |
| 2 | 6:00 – 12:00 | 5 minutes 5% - 9% | 30 seconds | 30 seconds |
| 3 | 12:00 – 18:00 | 5 minutes 10% - 14% | 30 seconds | 30 seconds |
| 4 | 18:00 – 24:00 | 5 minutes 15% | 30 seconds | 30 seconds |
| 5 | 24:00 – 30:00 | 5 minutes 15% | 30 seconds | 30 seconds |

Table 2: Outcome Measures – Frontal Plane

| Hip | Knee |
|---|---|
| Hip angle at initial contact (°) | Knee angle at initial contact (°) |
| Maximum hip angle before maximum knee flexion (°) | Maximum knee angle before maximum knee flexion (°) |
| Hip angle displacement (°) | Knee angle displacement (°) |
| Maximum internal hip moment (first 100 ms of ground contact) (N·m/(kgxm)) | Maximum internal knee moment (first 100 ms of ground contact) (N·m/(kgxm)) |

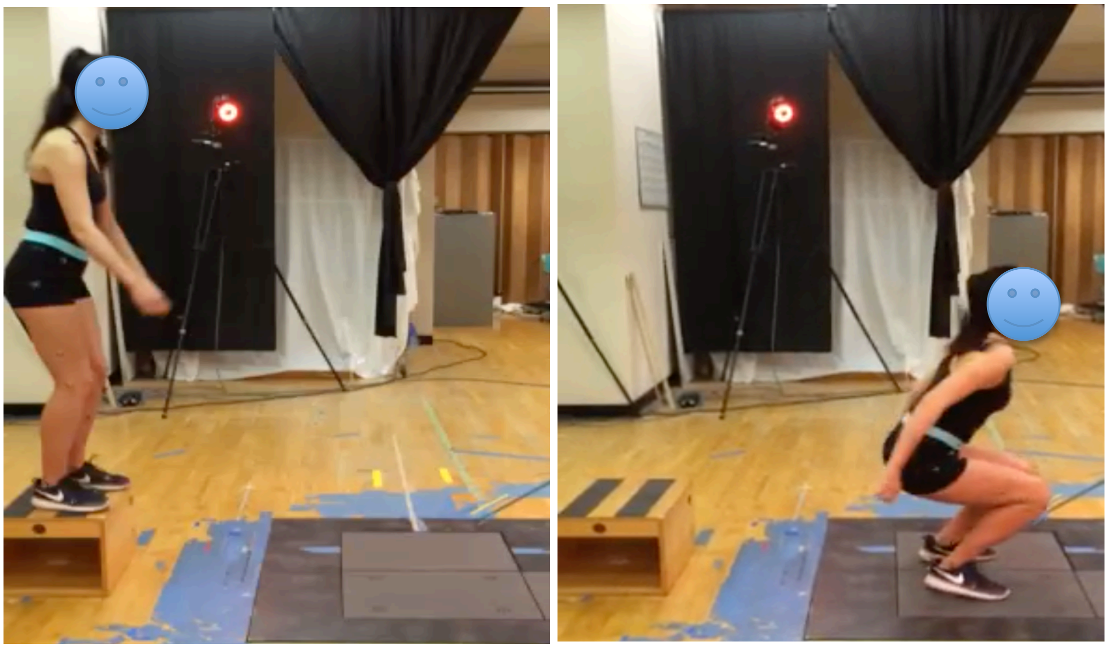

Figure 1. Drop jump landing task

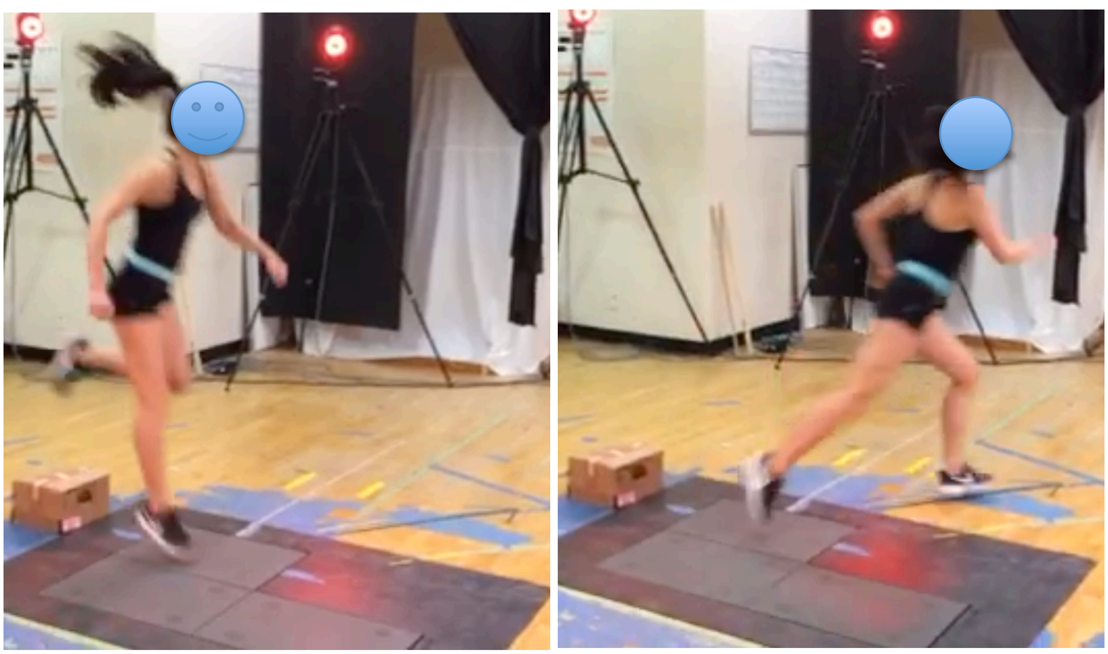

Figure 2. Side-step cut landing task

# CHAPTER 4

## Results

Tables 3 and 4 provide the means and standard deviations for each of the variables. There were no significant interactions for any of the variables (See Tables 3 and 4 and Figures 3-10). Additionally, there were no significant main effects for time for any of the variables (See Tables 3 and 4 and Figures 3-10). However, there were significant main effects for the task for hip angle at initial contact, hip angle displacement, and hip abduction moment.

For hip angle at initial contact, the drop jump task (pre-exercise: -7.04 ± 3.61°; post-exercise: -6.84 ± 3.55°) produced less hip abduction in the subjects than the side-step cut landing task (-11.31 ± 3.99°; -11.73 ± 4.89°). For frontal plane hip displacement, the drop jump task (3.03 ± 2.68°; 3.52 ± 3.02°) showed more abduction than the side-step cut landing task (5.54 ± 3.52°; 4.78 ± 3.63°). For hip abduction moment, the drop jump task had lower normalized internal moments (-0.08 ± 0.11 N·m/kgxm; -0.08 ± 0.09 N·m/kgxm) than the side-step cut landing task (-0.81 ± 0.18 N·m/kgxm; -0.75 ± 0.20 N·m/kgxm). The results are summarized below in Tables 3 and 4 and Figures 3-10.

Table 3. Frontal Plane Hip Mechanics

| Outcome Measure | Task | Time Pre | Time Post | |
|---|---|---|---|---|
| Hip angle at initial contact (°) | DJ | -7.04 ± 3.61 | -6.84 ± 3.55 | ** |
| | SSC | -11.31 ± 3.99 | -11.73 ± 4.89 | |
| Maximum hip angle (°) | DJ | -4.01 ± 4.76 | -3.32 ± 4.65 | |
| | SSC | -5.77 ± 5.16 | -6.95 ± 5.60 | |
| Hip angle displacement (°) | DJ | 3.03 ± 2.68 | 3.52 ± 3.02 | * |
| | SSC | 5.54 ± 3.52 | 4.78 ± 3.63 | |
| Hip abduction moment (N·m/kgxm) | DJ | -0.08 ± 0.11 | -0.08 ± 0.09 | *** |
| | SSC | -0.81 ± 0.18 | -0.75 ± 0.20 | |

(-) abduction/ (+) adduction; Main effect for task: *p=0.032, **p=0.006, ***p<0.001

Table 4. Frontal Plane Knee Mechanics

| Outcome Measure | Task | Time Pre | Time Post |
|---|---|---|---|
| Knee angle at initial contact (°) | DJ | -1.85 ± 2.90 | -2.09 ± 2.88 |
| | SSC | -2.49 ± 3.33 | -2.71 ± 3.20 |
| Maximum knee angle (°) | DJ | -6.32 ± 6.18 | -7.26 ± 5.79 |
| | SSC | -5.69 ± 5.26 | -6.67 ± 5.33 |
| Knee angle displacement (°) | DJ | -4.47 ± 4.89 | -5.17 ± 5.06 |
| | SSC | -3.46 ± 3.25 | -3.96 ± 3.50 |
| Knee adduction moment (N·m/kgxm) | DJ | 0.17 ± 0.13 | 0.18 ± 0.12 |
| | SSC | 0.08 ± 0.10 | 0.09 ± 0.11 |

(-) abduction/ (+) adduction

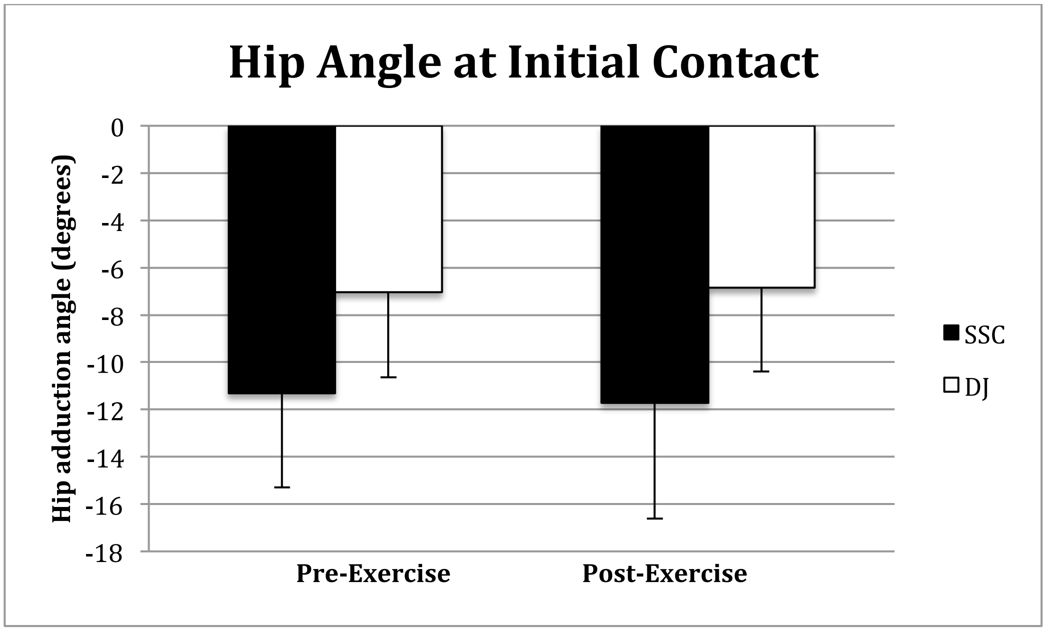


Figure 3. Hip adduction angle at initial contact. (-) abduction/ (+) adduction
There were no significant interactions (p=0.584) and no main effect for time (p=0.316). There was a significant main effect for task (p=0.006).

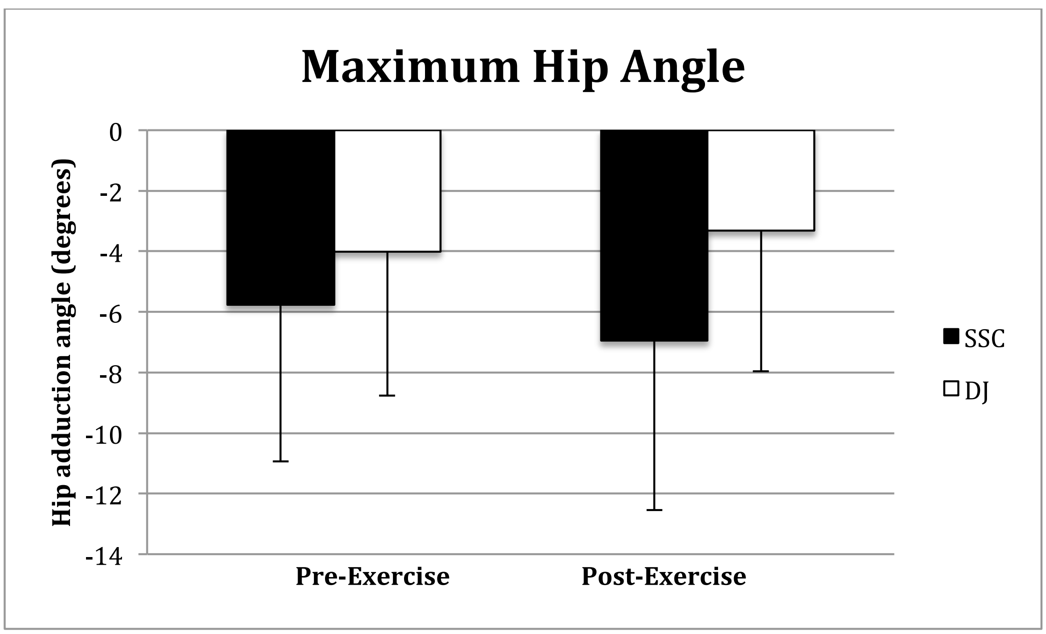


Figure 4. Maximum hip adduction angle. (-) abduction/ (+) adduction
There were no significant interactions (p=0.922), no main effect for time (p=0.447), and no main effect for task (p=0.362).

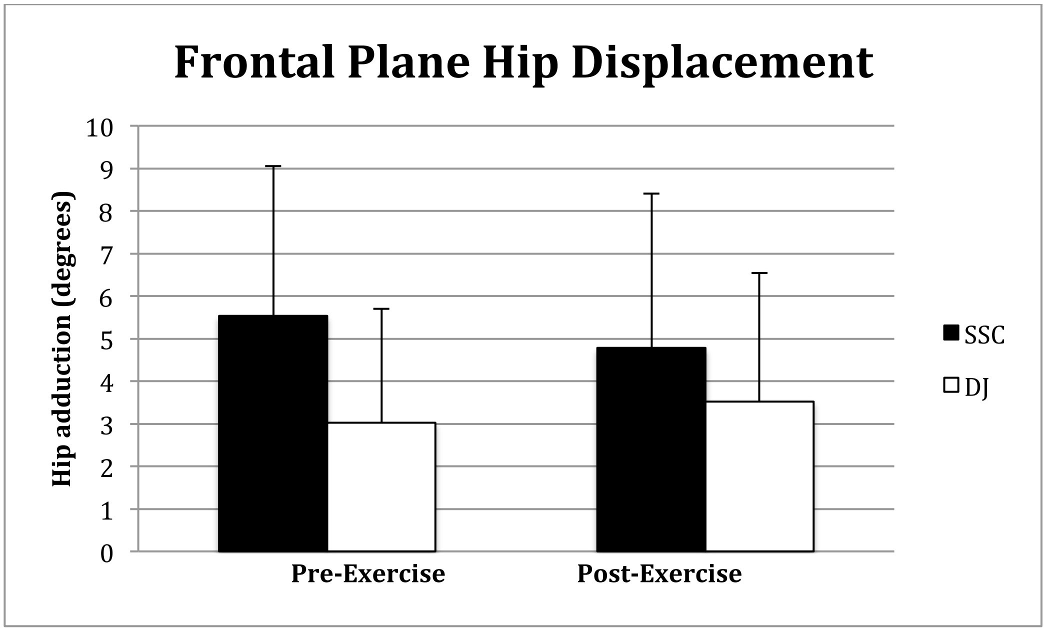


Figure 5. Hip angle displacement. (-) abduction/ (+) adduction
There were no significant interactions (p=0.460) and no main effect for time (p=0.954). There was a significant main effect for task (p=0.032).

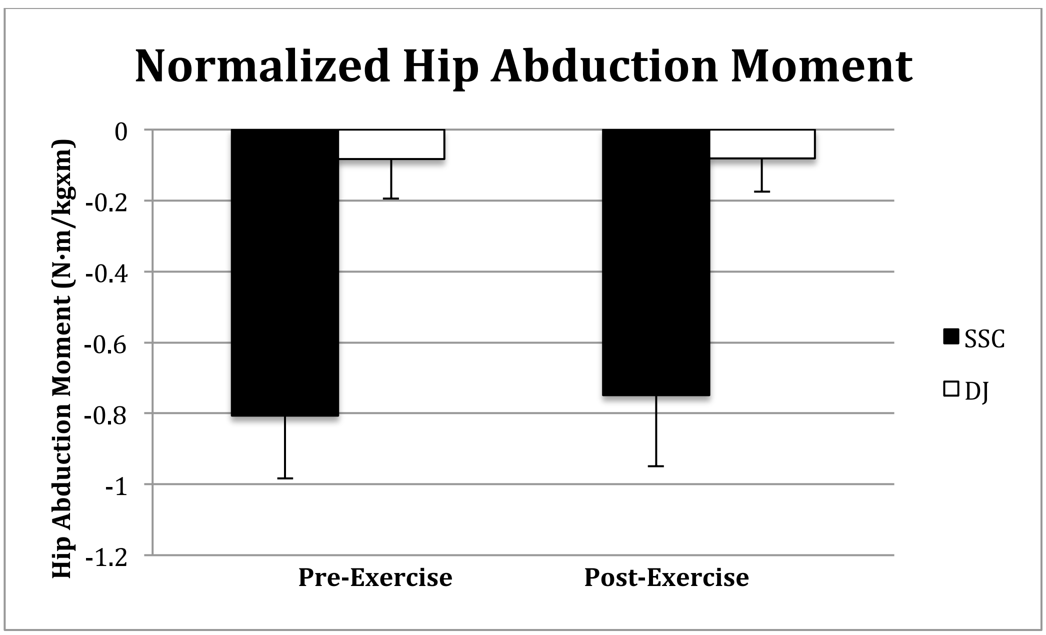


Figure 6. Internal hip abduction moment (normalized by height and weight).
(-) abduction/ (+) adduction. There were no significant interactions (p=0.886) and no main effect for time (p=0.810). There was a significant main effect for task (p<0.001).

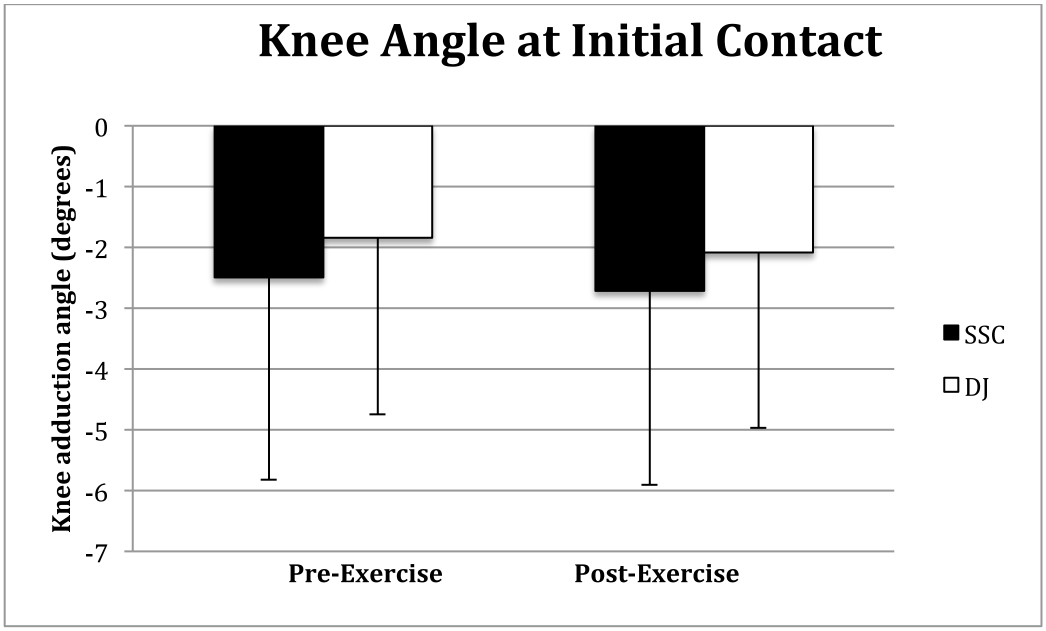


Figure 7. Knee adduction angle at initial contact. (-) abduction/ (+) adduction
There were no significant interactions (p=0.861), no main effect for time (p=0.304), and no main effect for task (p=0.940).

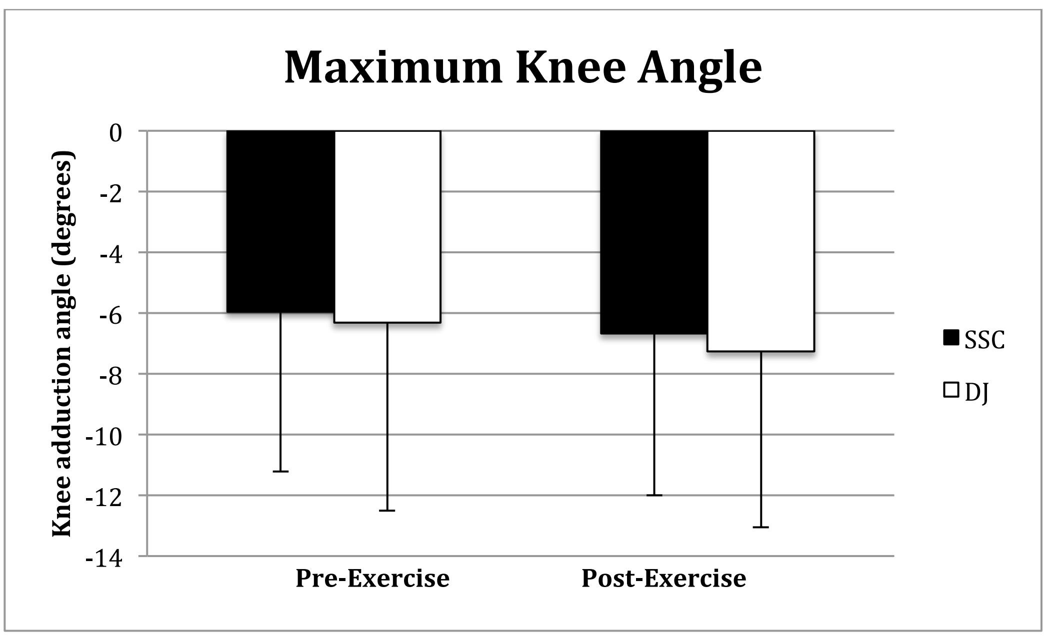


Figure 8. Maximum knee adduction angle. (-) abduction/ (+) adduction
There were no significant interactions (p=0.952), no main effect for time (p=0.215), and no main effect for task (p=0.885).

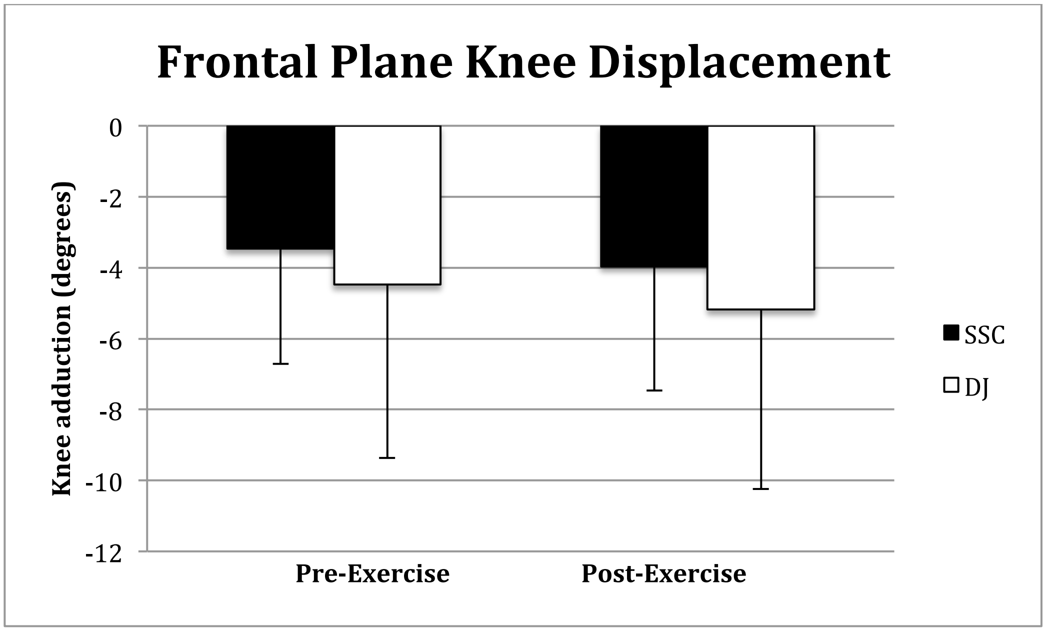


Figure 9. Knee angle displacement. (-) abduction/ (+) adduction
There were no significant interactions (p=0.955), no main effect for time (p=0.381), and no main effect for task (p=0.800).

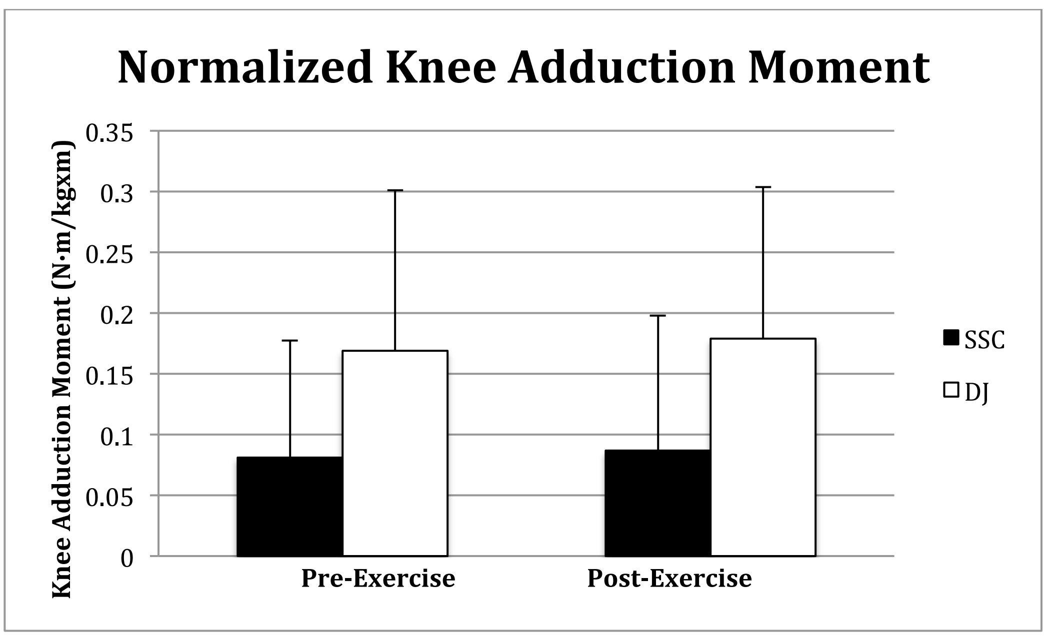


Figure 10. Internal knee adduction moment (normalized by height and weight). (-) abduction/ (+) adduction. There were no significant interactions (p=0.644), no main effect for time (p=0.773), and no main effect for task (p=0.051).

# CHAPTER 5

## Discussion

The purpose of this study was to determine the effects of a bout of exhaustive exercise on frontal plane knee and hip mechanics during a double-leg drop jump task and a single-leg side-step cut task. It was hypothesized that the exhaustive exercise protocol in this study would affect the frontal plane mechanics of the knee and hip in a way that greater stress was placed on the ACL, increasing the likelihood of an ACL injury. This hypothesis was supported by previous literature explaining that fatigue can significantly alter biomechanics by reducing both muscle force production and joint control (Cortes et al., 2012; Cortes et al., 2014). With reduced joint control it would be likely that one would see greater joint motion, allowing for a change in frontal plane mechanics of the knee. The prospective study by Hewett et al. (2005) showed that greater frontal plane knee motion, particularly an increased knee abduction angle and external knee abduction moment, was a strong predictor for an ACL rupture in the future. However, the primary finding of this study was that exhaustive exercise did not appear to have an effect on frontal plane hip and knee mechanics for either the drop-jump landing or the side-step cut landing. However, the side-step cut did have greater hip abduction at initial contact, greater overall hip displacement, and greater hip abduction moment than the drop-jump before and after exercise.

Contrary to the hypothesis, the exercise protocol in this study did not cause any effects on the frontal plane mechanics thought to place the ACL at risk of injury.

These findings did not align with the results of Shultz et al (2015), who found that greater frontal plane knee laxity and greater knee valgus motion were present across intermittent bouts of exercise that were meant to imitate a soccer match. Specifically, it involved six 12-minute bouts of shuttle runs that varied in distance and intensity with a 20-minute ‘halftime’ after three bouts. After each 12-minute bout, knee mechanics were tested using drop jumps, countermovement jumps, and perturbation trials. This differed from the current study’s protocol in that testing took place intermittently throughout the protocol versus just at the end. This testing directly after each bout of exercise could have provided better insight into the acute effects of each fatiguing bout. One possible explanation for the lack of effects in the current study between the exercise protocol and alterations in frontal plane knee mechanics is that the exercise protocol did not produce enough fatigue to change the mechanics during the post-test. This could be due to two possible reasons.

The exercise protocol was chosen because it had previously been shown by Chang et al. (2014) to produce effects of fatigue. In that study they reported a significantly decreased knee extension torque the day following completion of this protocol. While the protocol has been shown to reduce knee extension torque, it may not have specifically targeted the muscles necessary to produce dynamic stabilization in the frontal plane. The muscles most likely to contribute to frontal plane stabilization are the gluteal muscles (Hollman et al., 2012). Whereas, the muscles involved in producing the knee extension torque would likely have greater effects on the mechanics of the knee and hip in the sagittal plane. In this protocol, the gluteal muscles would have been targeted to some extent during the walking at an incline, 30 seconds of tuck jumping, and 30 seconds of lateral leaping, but this could possibly not have been enough stress to induce the fatigue-altered biomechanics mentioned in the literature.

Further, the current results are different than the Cortes et al. (2014) study that found a greater knee abduction angle at initial contact, coupled with significantly greater knee flexion, as subjects became more fatigued. They also reported hip abduction angle at initial contact, hip adduction moment at initial contact, and knee adduction moment at initial contact were not significantly different post- fatiguing protocol. The key difference between the present study and Cortes et al (2104) was greater use of specific fatiguing movements including vertical jumps, parallel squats, a 5-10-5 agility drill, and step-ups on a 30-centimeter box at a 200 step per minute pace. In addition, Cortes et al. (2014) objectively defined fatigue as less than 90 percent of their recorded maximum on the vertical jumps and similarly to Shultz et al. (2015) recorded biomechanical data from testing after each bout of exercise. Again, the intermittent testing of mechanics could have resulted in a larger fatigue response, leading towards greater changes in the mechanics.

Another potential reason that the exercise protocol may not have produced sufficient fatigue to alter frontal plane biomechanics during the post-testing could have been time between exercise and post-test jumping trials allowing for recovery from the exercise protocol. During this time, the reflective markers that had fallen off during the exercise protocol due to sweat were reapplied, the EMG wires that were used in the protocol but not in this study were reattached, and a static trial for marker calibration was repeated, all before the post-exercise drop-jump landing and side-step cutting trials were completed. While the transition time was not recorded for the first 20 subjects, for the last 20 subjects it took an average of $9.24 \pm 2.07$ minutes to finish the post-exercise trials. This is likely too long to wait to measure the acute effects of fatigue.

The second primary finding of the current study was that the two tasks appeared to have differential effects on frontal plane mechanics. Specifically, the side-step jump task had greater hip abduction angle at initial contact, overall hip angle displacement, and hip abduction moment irrespective of time. There were no main effects between the two tasks in the frontal plane mechanics of the knee. The lack of main effects on knee mechanics agrees with Kristianslund and Krosshaug (2013), who suggested that the drop vertical jump task, similar to the one used in the current study, was not sufficient to produce significant KAM. Yet, in the same light, the current study's results do not agree with their findings of significantly increased knee abduction angles both at initial contact and maximal knee abduction in the side-step cut when compared to a drop jump. However, they reported a side-step cut that involved less forward momentum and a much greater angle for the side-step cut than the one described in this study. That difference in angle could describe the difference in results. Variation between different cutting tasks was reported by Sigward et al. (2015) in relation to two side-step cuts, one at 45 degrees and one at 110 degrees from the initial plane of movement. They reported knee abduction moments were larger during the 110 degree side-step cut than during the 45 degree side-step cut. This different cutting task could explain the significant findings in the frontal plane knee mechanics by Kristianslund and Krosshaud that were not present in the current study.

Due to the lack of fatiguing effects and the presence of no significant interactions between task or time, further research is recommended in order to determine the extent of fatigue elicited in particular muscle groups. Another concern from the current study is that during data collection there were obvious differences between athletic abilities of the subjects participating in the study. While the inclusion criteria required the subjects to participate in 150 minutes of moderate to vigorous physical activity each week with some form of cutting or jumping sport in the last 6 months, it became apparent that this still left a large spectrum of females to participate in the study. As ACL injuries occur in many different athletic levels, a representation of all athletic abilities is ideal, but when results are combined, the varying skill sets between these two populations could impact the results. It could be hypothesized that those who are more athletic are stronger or more coordinated in their movements leading to less biomechanical alteration as a result of the exhaustive exercise. Further research could perform a stratification of participants into groups based on their amount of experience in cutting and jumping sports to hopefully create separation between the varied groups of athletic skill. Also, improvements to the study design would ideally produce a protocol that has no transition time between the fatiguing exercise and the post-exercise biomechanical testing. If the fatiguing protocol had produced fatigue in the subjects of this study, the lack of significant interactions could indicate that the exercise was targeting a different muscle than was necessary to alter mechanics in the frontal plane. This would lead to greater knowledge of which muscles dynamically control frontal plane motion during side-step cut and drop jump landings tasks. As frontal plane motion is known to increase ACL injury risk, identifying the factors that control frontal plane motion could lead to identification of ways to limit ACL injury risk by reducing frontal plane motion.

A particular limitation to this study is that when discussing dynamic movements that placed the ACL at greater risk of rupture, the focus was placed solely on the frontal plane. It must also be recognized that most ACL injuries occur during multiplanar movements that incorporate angles and moments in the sagittal and transverse planes as well. As shown by Cortes et al. (2014), a functional fatigue protocol can alter mechanics both in the sagittal and frontal planes. Further studies could be completed in regards to the interactions between motion in the different planes and the forces placed on the ACL.

# CHAPTER 6

## Conclusion

This study found hip angle at initial contact, hip angle displacement, and hip abduction moment between the side-step cut and drop jump landing tasks. However, there were no effects from the exercise, nor were there any significant interactions between the exercise and the two tasks. The lack of significant interactions and main effects could be due to the fact that the subjects were not sufficiently fatigued at the time of the post-exercise biomechanical test to alter the mechanics of either of the landing tasks.

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# APPENDIX

ANOVA Tables

Frontal Plane Hip Angle at Initial Contact

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 126.4 | 126.37 | 7.814 | 0.00601 |
| Time | 1 | 16.4 | 16.36 | 1.012 | 0.31644 |
| Task:Time | 1 | 4.9 | 4.87 | 0.301 | 0.58413 |
| Residuals | 124 | 2005.3 | 16.17 | | |

Maximum Frontal Plane Hip Angle Prior to Maximum Knee Flexion

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 21 | 21.455 | 0.836 | 0.362 |
| Time | 1 | 15 | 14.954 | 0.583 | 0.447 |
| Task:Time | 1 | 0 | 0.003 | 0.000 | 0.992 |
| Residuals | 124 | 3181 | 25.655 | | |

Frontal Plane Hip Displacement

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 43.7 | 43.68 | 4.694 | 0.0322 |
| Time | 1 | 0.0 | 0.03 | 0.003 | 0.9536 |
| Task:Time | 1 | 5.1 | 5.10 | 0.548 | 0.4604 |
| Residuals | 124 | 1153.9 | 9.31 | | |

Frontal Plane Knee Angle at Initial Contact

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 0.1 | 0.054 | 0.006 | 0.940 |
| Time | 1 | 10.3 | 10.258 | 1.066 | 0.304 |
| Task:Time | 1 | 0.3 | 0.298 | 0.031 | 0.861 |
| Residuals | 124 | 1192.8 | 9.620 | | |

Maximum Frontal Plane Knee Angle Prior to Maximum Knee Flexion

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 1 | 0.60 | 0.021 | 0.885 |
| Time | 1 | 45 | 44.82 | 1.551 | 0.215 |
| Task:Time | 1 | 0 | 0.10 | 0.004 | 0.952 |
| Residuals | 124 | 3584 | 28.90 | | |

Frontal Plane Knee Displacement

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 1.0 | 1.02 | 0.065 | 0.800 |
| Time | 1 | 12.0 | 12.20 | 0.772 | 0.381 |
| Task:Time | 1 | 0.0 | 0.05 | 0.003 | 0.955 |
| Residuals | 124 | 1958.1 | 15.79 | | |

Normalized Internal Knee Varus Moment (Normalized by height and body weight)

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 0.0412 | 0.04122 | 3.891 | 0.0508 |
| Time | 1 | 0.0009 | 0.00089 | 0.084 | 0.7726 |
| Task:Time | 1 | 0.0023 | 0.00227 | 0.214 | 0.6443 |
| Residuals | 124 | 1.3134 | 0.01059 | | |

Normalized Internal Hip Abduction Moment (Normalized by height and body weight)

| | Df | Sum Sq | Mean Sq | F value | Pr(>F) |
|---|---|---|---|---|---|
| Task | 1 | 3.625 | 3.625 | 152.270 | <2e-16 |
| Time | 1 | 0.001 | 0.001 | 0.058 | 0.810 |
| Task:Time | 1 | 0.000 | 0.000 | 0.021 | 0.886 |
| Residuals | 124 | 2.952 | 0.024 | | |