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Anterior cruciate ligament injuries are common in a wide variety of sports and most frequently occur during activities requiring rapid deceleration such as landing and cutting. Deceleration of the body's center of mass during movement results primarily from eccentric muscle contraction. This type of contraction serves to absorb energy from the whole-body system with the magnitude of energy absorbed directly related to the internal joint moment and the joint angular velocity. There is substantial evidence to demonstrate that following exercise females land with lesser knee flexion which may increase their risk for ACL injury. However, it is not known whether this change in knee position is a compensatory mechanism to overcome a reduction in quadriceps moment producing capacity that occurs during sustained exercise. It is possible that a more extended knee position is used after exercise in order to allow for greater knee flexion angular velocity so that the magnitude of knee energy absorption (EA) during landing can be maintained. Therefore, the purpose of this study was to: 1) evaluate the influence of exercise on the magnitude of knee EA during a single-leg jump-cut and, and 2) identify whether exercise influences the individual biomechanical determinants (internal knee moment and knee angular velocity) of knee EA. Forty recreationally active females performed single-leg jump-cuts before and after a standardized 30-minute exercise protocol. From recorded motion capture and ground reaction force data, the magnitude of knee EA, mean internal knee extension moment, and mean knee flexion angular velocity during the initial 100 milliseconds of landing were calculated. Despite no change in knee flexion angle at initial contact, females landed with 10% lesser knee EA following the exercise protocol. The lesser EA was the result of a 14% reduction in mean internal knee extension moment coupled with an 8% reduction in mean knee flexion angular velocity post-exercise. The results suggest that females utilized a stiffer landing strategy with lesser knee EA after sustained exercise. While the magnitude of the EA reduction observed during the single-leg jump-cut is probably not clinically meaningful, it is likely that the adoption of a stiffer landing strategy following exercise during more demanding movement tasks might result in increased loading of static structures and greater lower extremity injury risk.

The Effect of Exercise on the Biomechanical Determinants of Knee Energy Absorption during Single-leg Jump-Cuts.

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

Anne E. Pollard, Author

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CHAPTER 1

Introduction

1.1 Introduction

Anterior cruciate ligament (ACL) injuries are common (Griffin et al., 2000) in a wide variety of sports (Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Toth & Cordasco, 2001), but especially in those that involve frequent changes of direction, pivoting, and cutting maneuvers (Griffin et al., 2000). These injuries are usually sustained through a non-contact mechanism (Jacobs et al., 2007) and are more likely to occur in younger athletes (Griffin et al., 2000) and females (Hewett, 2005; Toth & Cordasco, 2001).

ACL injuries are expensive (Mather, 2013), and do not always result in positive outcomes, with up to 16.0% of ACL reconstructed patients sustaining a second ACL injury to either the contralateral or ipsilateral knee within 10 years of the initial injury (Chahal, Lee, Heard, & Bach, 2013; Leroux et al., 2014; Paterno, Rauh, Schmitt, Ford, & Hewett, 2012). Further, 65% of ACL injuries are accompanied by a concomitant meniscus tear (Wyatt, Inacio, Liddle, & Maletis, 2013), which increases the risk for early-onset osteoarthritis (Butler, Minick, Ferber, & Underwood, 2009; von Porat, 2004). Multiple studies have reported osteoarthritic changes to be present in individuals within 5 to 15 years after sustaining an ACL injury (Li et al., 2011; von Porat, 2004), which dramatically increases the likelihood of needing a total knee arthroplasty later in life (Van Manen, Nace, & Mont, 2012). Further, individuals can suffer from intense psychological and emotional difficulties following an ACL injury event (Ardern et al., 2014; Langford, Webster, & Feller, 2009). As a result, it is imperative to investigate means that could assist in the prevention of these devastating injuries.

While factors such as knee anatomy and hormonal influences on tissue function are important to consider when investigating ACL injuries, biomechanical risk factors have received significant attention because they are considered to be modifiable. Specific to the sagittal plane, an eccentric quadriceps contraction has been shown to induce anterior tibial shear force (ATSF) that can strain the ACL (DeMorat, 2004). During landing and cutting tasks, the body must produce internal extension moments in order to decelerate the whole body center of mass. At the knee, this resultant internal knee extension moment (KEM) during landing and cutting is driven primarily by eccentric contraction of the quadriceps, with a large magnitude KEM considered detrimental to ACL loading, particularly when the knee is in a less flexed position (DeMorat, 2004). The reason for this is that during quadriceps contractions at decreased knee flexion angles, a greater component of the quadriceps force is directed anteriorly on the tibia (ATSF) with high quadriceps loading shown to cause a large anterior displacement and ligamentous disruption and/or full ACL rupture in cadaveric knees (DeMorat, 2004). This increase in ATSF at lesser knee flexion angles secondary to a given magnitude of quadriceps loading has been attributed to an increase in the patella tendon-tibia shaft angle (Yu & Garrett, 2007), which is greatest when the knee is in full extension (Nunley, Wright, Renner, Yu, & Garrett, 2003). Further, the elevation angle of the ACL also increases as the knee moves into lesser flexion, which results in a greater proportion of the loading on the ACL due to ATSF being shear compared to tensile (Blackburn &

Padua, 2008). As ligament is more resistant to tensile forces, the ACL is at a greater risk for injury due to the large amount of shear loading present during high magnitude quadriceps contractions in less flexed knee positions (Woo, Hollis, Adams, Lyon, & Takai, 1991).

There is substantial evidence to demonstrate that exercise induces changes in lower extremity landing biomechanics (Chappell, 2005; Cortes, Greska, Kollock, Ambegaonkar, & Onate, 2013; Kernozek, Torry, & Iwasaki, 2007) and that these changes might increase the risk for ACL injury (Chappell, 2005; Cortes et al., 2013; Kernozek et al., 2007). After exercise, subjects have generally been shown to land with lesser knee flexion angles (Chappell, 2005; Kernozek et al., 2007; Lucci, Cortes, Van Lunen, Ringleb, & Onate, 2011). However, the results pertaining to sagittal plane knee kinetics are less consistent. Chappell et al (2005) reported increased ATSF after exercise in females performing a stop-jump task, while Kernozek et al (2007) identified decreased ATSF after exercise in females performing terminal drop landings. Further, while exercise generally results in a decrease in resultant KEM (Chappell, 2005; Kernozek et al., 2007; Orishimo & Kremenic, 2006), subjects have been shown to utilize a greater range of motion to complete landing tasks after performing an exercise protocol (Shimokochi, Ambegaonkar, Meyer, Lee, & Shultz, 2013). Collectively, these kinematic and kinetic adaptations following exercise potentially allow individuals to successfully perform landings (i.e., deceleration of the whole body center of mass), but through different underlying mechanisms.

Eccentric muscle contraction occurs when the net internal joint moment (M) and the joint angular velocity (ω) are acting in opposite directions (Winter, 2005).

Though the kinetic energy of the body can be passively absorbed in the bones, ligaments, and articular cartilage, (Coventry, O'Connor, Hart, Earl, & Ebersole, 2006) the majority of energy is actively absorbed in the muscles via eccentric contractions during landing (Winter, 2005). At the knee, energy absorption (EA) via eccentric contraction from the quadriceps is extremely important for successfully decelerating the body with the knee serving as a major contributor to lower extremity energy absorption during landing (Winter, 2005). However, given that the magnitude of EA is driven by the combined influences of the net KEM and joint angular velocity, there is the potential that different joint kinematic and kinetic profiles can be used to achieve the same magnitude of EA (Norcross et al., 2014). Therefore, the same changes in landing strategy following exercise (i.e., lesser knee flexion at initial contact) that could increase injury risk might also allow individuals to maintain a similar magnitude of knee EA, but through different underlying mechanisms (Coventry et al., 2006; Norcross et al., 2014)

It has been hypothesized that following exercise, when the moment-producing capacity of the quadriceps is decreased, the magnitude of knee EA might be maintained by adopting a landing strategy that would reduce the mean KEM requirement while allowing for increased joint angular velocity (Norcross et al., 2014). To achieve this, individuals could land with lesser knee flexion, which might decrease their mean KEM requirement by positioning the ground reaction force vector closer to the knee joint center. However, this change in knee flexion angle at initial contact could also allow for greater joint angular velocity, thereby allowing individuals to maintain the same magnitude of EA pre and post-exercise, despite a reduction in resultant KEM. Therefore, when the capacity to generate KEM is compromised, such as through a decrease in quadriceps force following exercise, it is plausible that individuals might utilize a more erect (decreased knee flexion) and higher risk landing posture in order to maintain the magnitude of knee EA necessary to successfully complete the landing task (Norcross et al., 2014).

Given this possibility, the purpose of this study was to: 1) evaluate the influence of exercise on the magnitude of knee EA during a single-leg land and cut task, 2) identify whether exercise influences the individual biomechanical determinants (internal knee extension moment and angular velocity) of knee EA, and 3) evaluate the influence of exercise on knee flexion angle at initial contact during a single-leg jump-cut task. We hypothesized that after exercise: 1) the magnitude of energy absorption would not change after exercise, but that 2) individuals would exhibit lesser internal knee extension moment and greater knee flexion angular velocity, and 3) lesser knee flexion at initial contact.

1.2 Research Question

Research Question 1: What is the influence of exercise on the magnitude of knee EA during a single-leg jump-cut task?

Research Hypothesis 1: The magnitude of knee EA will not change after exercise.

Research Question 2: What is the influence of exercise on the individual biomechanical determinants (internal knee extension moment and angular velocity) of knee EA during a single-leg jump-cut task?

Research Hypothesis 2: After exercise, individuals will exhibit lesser internal knee extension moment and greater knee flexion angular velocity.

Research Question 3: What is the influence of exercise on knee flexion angle at initial contact during a single-leg jump-cut task?

Research Hypothesis 3: After exercise, individuals will exhibit lesser knee flexion angles at initial contact.

1.3 Operational Definitions

Initial ground contact: The beginning of the landing period was defined as the instant when the vertical component of the ground reaction force exceeds 10 Newtons.

Dominant leg: Subjects' leg dominance was then assessed as described by Hoffman et al. (1998). In short, patients performed three tests to determine limb dominance. Dominant limb tests included instructing the participant to kick a ball to the assessor, to step up onto a step, and to recover after a posterior perturbation. The dominant limb was defined as the limb used at least twice to kick the ball, step up first, and utilized to regain balance.

Single-leg jump-cut: Participants were instructed to jump off both legs over a 17cm high hurdle, land with their dominant foot positioned in the center of the force plate, and cut at a 60 degree angle as quickly as possible in the direction opposite of their dominant leg (e.g., right leg dominant participants will cut toward the left).

1.4 Assumptions & Limitations

The following assumptions were made:

1. Participants performed all testing protocols with maximal effort and to the best of their ability.
2. Participants were honest regarding their previous history with respect to the inclusion/exclusion criteria.
3. The biomechanical data that was collected during this experiment was reliable and is valid for all participants.

1.5 Delimitations

The following delimitations were made:

1. All participants were between the ages of 18-30 at the time of testing.
2. All kinematic and kinetic data was sampled using the same motion analysis system and force plates.
3. All participants reported no history of lower extremity injury, lower extremity surgery, ACL injury, or neurological disorder that restricted activity for more than 3 days within the 6 months prior to data collection.
4. All participants reported being physically active as defined by participation in at least 150 minutes of moderate to vigorous physical activity per week.

CHAPTER 2

Literature Review

2.1 Introduction

ACL injuries have been described as being multifactorial and therefore require investigation into multiple areas in order to understand their significance, mechanism, and to develop ways to prevent these debilitating injuries. While it is not necessary to thoroughly review all research related to ACL injuries, several factors are important to examine in order to provide context for the proposed research study. Some of these factors include: injury epidemiology, sagittal plane biomechanical risk factors for injury, the effect of exercise on sagittal plane biomechanics, lower extremity energy absorption, and the effect of exercise on lower extremity energy absorption. Through the review of previous literature on these topics, this review supports the importance of the investigation that was conducted.

2.2 ACL Injury Epidemiology

ACL injuries are common in athletes between the ages of 15 and 25 and commonly occur during participation in sports involving frequent changes of direction, pivoting, and cutting maneuvers (Griffin et al., 2000). Approximately 200,000 ACL tears occur every year (Center for Disease Control and Prevention, 1996), with upwards of 175,000 of these injuries resulting in ACL reconstruction surgery (Lyman, 2009). The average cost to society for lost work, earnings, disability, knee osteoarthritis expenses, and total knee arthroscopy secondary to primary ACL injuries is estimated to be $7.6 billion annually (Mather, 2013).

Further, the health outcomes of the injury and subsequent surgery and/or rehabilitation are not always positive (Brophy et al., 2012; Faltstrom, Hagglund, & Kvist, 2013). Many ACL-injured athletes experience a decreased quality of life due to early-onset osteoarthritis, fail to return to the same level of play in their respected sport, and are at increased risk of re-injury (Bahr & Krosshaug, 2005; Brophy et al., 2012).

2.2a Injury Occurrence

Approximately 200,000 ACL tears occur in the United States every year (Center for Disease Control and Prevention, 1996), with the majority of these tears occurring in athletes (Ireland, 1999). Approximately 70% of ACL injuries are non-contact in nature and occur during a cutting or pivoting maneuver without direct contact or a collision with another player (Jacobs et al., 2007). Non-contact ACL injuries typically occur when the athlete is trying to change direction quickly or when landing from a jump and can be generalized as deceleration injuries (Griffin et al., 2000; Toth & Cordasco, 2001). While ACL injuries occur in both males and females, females have a significantly higher incidence rate of ACL tears when compared to males (Hewett, 2005; Toth & Cordasco, 2001). It has been reported that females are at least 4 times more likely to injure their ACL compared to their male counterparts participating in similar types of activities (Arendt & Dick, 1995). Additionally, it has been found that specific sports tend to have a higher rate of ACL injuries in the female athletic population. Females participating in volleyball, basketball, soccer, rugby, and gymnastics are typically at greatest risk for ACL injury (Jacobs et al., 2007; Toth & Cordasco, 2001). All of these sports involve deceleration, cutting, and pivoting, which have been established as common mechanisms for ACL injury. As demonstrated, ACL injuries are relatively common and remain a health concern given the large number of injuries that occur each year (Lyman, 2009) and the increasing number of females that are participating in sporting activity (Hootman, Dick, & Agel, 2007).

2.2b Cost of ACL Reconstruction Surgery

It has been estimated that one ACL surgery costs a patient approximately $17,000 (Toth & Cordasco, 2001). However, the initial surgery cost is only one factor contributing to the total financial costs related to ACL injury. The economic impact in the United States alone is estimated to be $7.6 billion annually, with this amount encompassing wages from lost work, earnings, disability, knee osteoarthritis, and future total knee arthroscopy (Mather, 2013). Surgery is also not the only direct immediate medical cost following an ACL tear, as a patient will typically have anywhere from 4-12 months of weekly rehabilitation following the procedure (Delay, Smolinski, Wind, & Bowman, 2001). The base fee for an hour of one-on-one time with a physical therapist in their office can cost upwards of $55 an hour (Bureau of Labor Statistics, 2012), with patients being seen between two and three times per week. Even after the entire process is complete, there is still the chance for subsequent surgeries and complications, as one reconstruction surgery does not guarantee that the reconstructed tissue or the opposite ACL will not fail.

Shelbourne et al (2009) reported that after an initial ACL surgery, approximately 5.3% of subjects had contralateral ACL ruptures, and 4.3% of individuals had ipsilateral recurrent ruptures. Additionally, Leroux et al (2014) reported that 3.4% of subjects had a contralateral ACL reconstruction. Other studies have reported that the rate of contralateral ACL injury ranges from 7.5% to 16.0%, which is approximately 32,000 additional surgeries (Chahal et al., 2013; Wasserstein et al., 2013; Wright, Magnussen, Dunn, & Spindler, 2011).

Not only are second injuries and surgeries occurring, but these incidences have also been found to be associated with age. Shelbourne et al (2009) reported that the rate for a subsequent ACL surgery was age dependent, with 17% of necessary surgeries occurring on patients younger than 18 years of age. Hettrich et al (2013) investigated the incidence of surgery secondary to initial anterior cruciate ligament reconstruction (ACLR) complications 6 years after the initial surgery. This study also found younger age to be a predictor for subsequent surgery and reported that 18.9% of patients that underwent an initial ACLR required another surgery on the ipsilateral knee (Hettrich et al., 2013). Since most ACL injuries occur in athletes ages 15 to 25 (Griffin et al., 2000), it is concerning that both Shelbourne et al (2009) and Hettrich et al (2013) reported that younger age was associated with a secondary surgery following the initial ACLR.

Not only has it been shown that ACLR patients are at risk to sustain a second injury, but Paterno et al (2012) compared the injury rate of ACLR patients to healthy individuals with no history of prior knee injury during the first 12 months after return to sport. Paterno et al (2012) investigated the incidence rate of a second ACL injury in either the ipsilateral or contralateral knee during the first 12 months after return to sport. The study reported that 16 individuals sustained a second ACL injury and only one subject in the referent group sustained an initial ACL injury (Paterno et al., 2012). From their results, the study concluded that ACLR patients have a 15-fold greater risk of a secondary ACL injury when compared to a healthy population (Paterno et al., 2012). Additionally, the study provided insight on the location of the second injury, reporting that 75% of the ACLR group sustained the tear in their contralateral knee (Paterno et al., 2012). The results of this study clearly demonstrate that patients who have already sustained an ACL injury with subsequent surgery are much more likely to sustain a second injury when compared to a healthy population (Paterno et al., 2012).

A second injury would likely require another surgery, and the more surgeries an athlete undergoes, the more likely it is for their quality of life to decrease. For a young athlete, a decrease in quality of life would typically be characterized by decreased mood, pain and functional limitations, and the athlete being unable to return to their previous level of competition in sport. Brophy et al (2012) investigated whether gender and age had a significant effect on initial return to play. The study found that younger males were more likely to initially return to play after an ACL reconstruction surgery. However, men were also more likely to attribute the reason they were no longer playing to their ACL injury (Brophy et al., 2012). It has already been established that females are known to suffer ACL injuries at an increased rate compared to males, so the lower rate of initial return to play in females is unsettling, as the frustrations that occur from injury could cause a decline in their quality of life. This decline could continue even further, because after ACL reconstruction surgery and rehabilitation, other problems may still occur (Butler et al., 2009; Li et al., 2011; Lohmander, Englund, Dahl, & Roos, 2007; von Porat, 2004).

While not all ACL-reconstructed knees undergo a second injury, it is known that individuals who have undergone ACLR are at an increased risk for osteoarthritis (OA) later in life (Li et al., 2011; von Porat, 2004). The Mayo Clinic defines OA as “the most common form of arthritis affecting millions of people worldwide…and occurs when the protective cartilage on the ends of your bones wears down over time. Osteoarthritis gradually worsens, and no cure exists” (Mayo Clinic, 2014). An even more concerning piece related to OA is that many ACL-injured individuals will develop this condition relatively early in life. Butler et al (2009) and von Porat et al (2004) concluded that ACL injury leads to early-onset OA, with the first symptoms occurring as soon as 5 to 15 years after the initial ACL injury (Roos, Adalberth, Dahlberg, & Lohmander, 1995). Von Porat et al (2004) conducted a study looking at knee OA 14 years after an ACL injury had occurred and found staggering results. Using radiographic imaging, approximately 80% of the subjects had significant radiographic changes in the knee (von Porat, 2004). Additionally, more than 40% of the subject population had radiographic changes that categorized them for definite OA, and the majority of these participants had subsequent knee pain that altered their quality of life (von Porat, 2004). Similarly, Li et al (2011) determined the prevalence of radiographic OA in ACL-reconstructed patients to be 38.6% at a median follow-up time of 7.35 years. As shown, ACL injuries precipitate an early-onset of OA resulting in patients typically complaining of stiffness, functional impairment, and varying levels of pain with daily activities (Lohmander et al., 2007).

The risk of OA is increased when a meniscal tear accompanies the ACL sprain (von Porat, 2004). Of the 200,000 ACL injuries that occur annually, it is estimated that 130,000 have concomitant meniscal tears (Wyatt et al., 2013). Surgeons typically repair a tear via resection of the meniscus, an approach that decreases the total amount of protective cartilage in the knee and further increases the likelihood for osteoarthritis. Since 65% of ACL injuries are not isolated (Wyatt et al., 2013), this debilitating consequence is important to consider, as an increased risk of osteoarthritis has been demonstrated for patients that experience more soft tissue damage (Lohmander et al., 2007). Further, osteoarthritis increases the likelihood that a total knee replacement will be necessary later in life (Van Manen et al., 2012).

In addition to the obvious physical difficulties that incur after experiencing an ACL injury and surgical repair, there are also intense emotional and psychological challenges (Ardern et al., 2014; Morrey, Stuart, Smith, & Wiese-Bjornstal, 1999). After surgery, it has been found that patients usually experience negative mood changes both at two weeks and two months after surgery, with varying mood changes throughout the entire rehabilitation process (Morrey et al., 1999). The rehabilitation process can be exhausting, but the end goal is always to return back to sport. Ardern et al (2014) concluded that psychological readiness was the most significantly associated factor when returning to sport. Additionally, the two most common reasons for not returning to sport were “lack of trust in the knee” and the “fear of sustaining a new injury”, which are both highly related to psychological wellness (Ardern et al., 2014). Langford et al (2009) found that the subjects who did return to sport reflected a much more positive psychological response than those who had not returned to their competitive sport. These investigations highlight that the ramifications following ACL injury are not limited to physical function. Rather, it is important to recognize and address psychological aspects as well throughout the rehabilitation process.

2.2c Conclusion

ACL injuries are expensive and do not always result in positive outcomes, as individuals are at an increased risk for a subsequent ACL injury to either the ipsilateral or contralateral side. ACL and any concomitant meniscal injury can also result in early onset osteoarthritis, which typically results in unfavorable consequences in the future and the potential need for a total knee arthroplasty. Additionally, intense emotional and psychological difficulties can hinder the rehabilitation process and prolong or prevent return to play. Due to the debilitating nature of these injures, it is important to determine means that will aid in prevention.

2.3 Risk Factors Related to ACL Injury

ACL injuries have been described as multifactorial, with anatomical (Boden, Sheehan, Torg, & Hewett, 2010; Uhorchak et al., 2003), hormonal (Hoffman, Harter, Hayes, Wojtys, & Murtaugh, 2008; Shultz, Sander, Kirk, & Perrin, 2005; Wojtys, Huston, Lindenfeld, Hewett, & Greenfield, 1998), and biomechanical factors thought to contribute to ACL injury risk (Alentorn-Geli et al., 2009; Hewett, 2005). The focus of this study will be on the modifiable biomechanical factors specifically related to sagittal plane landing mechanics. In the sagittal plane, internal knee extension moment (KEM) has been studied extensively in regards to its relationship to ACL injury risk (Boden, Dean, Feagin, & Garrett, 2000; Myer et al., 2009). Similarly, a considerable amount of work has been done to investigate the influence of knee flexion angle on ACL injury risk and loading (DeMorat, 2004; Markolf et al., 1995). The above two mentioned factors have been previously studied on numerous occasions and are regarded as exceptionally relevant to sagittal plane ACL loading and injury risk. Due to their importance, this study will investigate both sagittal plane factors.

2.3a Sagittal Plane Knee Biomechanics

It is well established that the primary purpose of the ACL is to prevent anterior translation of the tibia, and the primary action of the quadriceps musculature is to produce an internal extension moment at the knee. While eccentric quadriceps contraction is necessary for controlling knee flexion during landing, increased quadriceps force coupled with lesser knee flexion creates an unfavorable loading situation with respect to ACL injury risk.

An active quadriceps contraction produces an anteriorly directed force on the tibia due to its attachment site on the tibial tuberosity, which increases the strain on the ACL (DeMorat, 2004). During landing from a jump or cutting, the quadriceps contracts eccentrically to control knee flexion and attempts to slow down the body's center of mass (Winter, 2005). The resultant knee moment during this movement, which is driven by the quadriceps, is quantified as the internal knee extension moment (KEM), which is a relevant variable of interest in regards to sagittal plane knee biomechanics (Kernozek et al., 2007; Orishimo & Kremenic, 2006).

Boden et al (2000) reported that a vigorous eccentric quadriceps contraction may play an important role in disruption of the ACL, with this premise supported through research. DeMorat et al (2004) conducted a cadaveric study to investigate the ACL's response to an aggressive and isolated quadriceps load at 20 degrees of knee flexion. The study used a quadriceps load of 4500 N, which caused ligamentous disruption in 55% of knees and caused a complete ACL rupture in 27% (DeMorat, 2004). On average, the anterior tibial displacement caused by the strong quadriceps force was 19 mm (DeMorat, 2004). From their study, DeMorat et al (2004) suggested that the quadriceps muscle can serve as the "major intrinsic force in a noncontact ACL injury" and can increase the amount of strain placed on the ACL.

Markolf et al (1995) measured the direct force on the ACL under various loading conditions and knee flexion angles. The study determined that anterior tibial shear force resulted in markedly increased ACL elongation at hyperextension to 45 degrees of knee flexion compared to knee positions greater than 45 degrees (Markolf et al., 1995).

A lesser knee flexion position has been classified as an unfavorable position with respect to ACL loading (DeMorat, 2004; Markolf et al., 1995). Markolf et al (1995) discovered that an increase in anterior tibial shear force puts more strain on the ACL. It has also been established that under similar quadriceps loading conditions, lesser knee flexion angle results in greater anterior shear force on the knee, (DeMorat, 2004) with this increase in anteriorly directed force attributed to an increase in the patella tendon-tibia shaft angle (Yu & Garrett, 2007). Nunley et al (2003) found that females had the greatest patella tendon-tibia shaft angles at full extension, which would therefore result in greater anterior tibial shear force under similar quadriceps loading conditions in more extended vs. more flexed positions.

In addition, lesser knee flexion increases the ACL elevation angle (Herzog & Read, 1993). At greater angles, the ACL is oriented more vertically, so there is an increased amount of shear loading relative to tensile loading on the ACL (Yu & Garrett, 2007). Since the ACL is more resistant to tensile forces than shear forces, the ACL is at greater risk for failure during quadriceps loading in less flexed knee positions (Woo et al., 1991). With the knee closer to full extension, a greater proportion of the quadriceps force is directed anteriorly, with more of this force applying shear loading (Blackburn & Padua, 2008).

2.3b Conclusion

A strong quadriceps contraction increases the anterior pull on the tibia and places the ACL at an increased risk for injury. During landing and deceleration activities, the resultant internal knee extension moment is indicative of the magnitude of contraction of the quadriceps, with large magnitude KEM considered detrimental to ACL loading. It has also been reported that decreased knee flexion angles cause: 1) an increase in the patella tendon-tibia shaft angle, which increases the amount of quadriceps force that is directed anteriorly; and 2) an increase in the ACL elevation angle, which increases the amount of shear vs. tensile loading on the ACL. Cadaveric studies have demonstrated that increased quadriceps loads at lesser knee flexion angles increases the anterior tibial shear force, and can cause ligamentous disruption and/or full rupture of the anterior cruciate ligament in cadaveric specimens. Therefore, it can be concluded that a forceful eccentric quadriceps contraction coupled with decreased knee flexion is detrimental and unfavorable for the integrity of the ACL.

2.4 Effect of Exercise on Sagittal Plane Biomechanics

There is substantial evidence in the literature that demonstrates the changes exercise induces on lower extremity landing biomechanics (Chappell, 2005; Cortes et al., 2013; Cortes, Quammen, Lucci, Greska, & Onate, 2012; Kernozek et al., 2007; Lucci et al., 2011; Mclean et al., 2007; Schmitz et al., 2014). Typically, these biomechanical changes are considered detrimental and likely place the individual at an increased risk for injury (Chappell, 2005; Cortes et al., 2013; Kernozek et al., 2007).

Several different protocols have been used in studies to induce fatigue in individuals before performing a specific movement task. Cortes et al (2013) utilized a multivariate short-term fatigue protocol which consisted of countermovement jumps, step-ups and step-downs, squats, and a pro-agility shuttle run. This study not only looked at pre-fatigue and post-fatigue measures, but also at the time of 50% fatigue, which was defined as the halfway point in the total amount of sets performed (Cortes et al., 2013). Subjects were asked to perform 2 sets of unanticipated stop jumps and side steps after the exercise bout was completed (Cortes et al., 2013). It was reported that at 100% fatigue, peak knee flexion angles and knee flexion angles at initial contact were decreased when compared to the pre-fatigue and 50% fatigue measures (Cortes et al., 2013). Similarly, Cortes et al (2012) utilized the same fatigue protocol, but used both unanticipated stop jumps into vertical jumps and sidesteps as tasks. Similar results were reported, with significantly lesser peak knee flexion reported after exercise regardless of task (Cortes et al., 2012).

Lucci et al (2011) conducted a study utilizing a similar short-term fatigue protocol approach as Cortes et al (2013); but with some modifications to the individual exercises. Fatigue was induced through four sets of four different exercises: step-ups and step-downs, the L-drill, vertical jumps, and an agility ladder (Lucci et al., 2011). The entire exercise series was completed without any rest (Lucci et al., 2011). Outcome variables were measured during an unanticipated side-step cutting task, which was performed both prior to and after the exercise protocol (Lucci et al., 2011). As with most other studies, peak knee flexion and knee flexion angle at initial contact significantly decreased after exercise. However, the most dramatic finding in the study was that participants altered their knee mechanics after only 5 minutes of physical activity (Lucci et al., 2011). This finding suggests that not only does exercise induce biomechanical changes that are thought to be detrimental to ACL injury risk, but these changes can occur within a very short period of time (Lucci et al., 2011).

Chappell et al (2005) used vertical jumps and sprints to induce fatigue in subjects. Subjects who performed a double-leg stop jump task followed by a vertical jump before and after exercise (Chappell, 2005). Chappell et al (2005) reported that after exercise, females exhibited a decrease in knee extension moment, but an increase in anterior tibial shear force. In this study, the knee flexion angle was measured at the point of peak proximal tibial anterior shear force (Chappell, 2005). After exercise, knee flexion angle at peak ATSF significantly decreased, from 29.9 degrees to 25.7 degrees, representing a mean decrease of 14% (Chappell, 2005). Despite the decreased KEM after exercise, females exhibited lesser knee flexion angles and greater anterior tibial shear force than pre-fatigue, which likely increases the risk for ACL injury (Chappell, 2005).

Orishimo and Kremenic (2006) used step-ups to induce fatigue in subjects who performed a single-leg forward hop before and after exercise. This study reported that after exercise, peak knee extension moment decreased and there was a significantly greater total range of motion at the knee over the landing period, which was the most notable finding in the study (Orishimo & Kremenic, 2006). They proposed that the greater range of motion was due to the inability to quickly slow down their center of mass due to fatigue (Orishimo & Kremenic, 2006).

Kernozek et al (2007) utilized a fatigue protocol consisting of repeated squats and had subjects perform multiple single-leg drop landings before and after the protocol was completed. The results of the study showed landing profiles similar to proposed noncontact ACL mechanisms (Kernozek et al., 2007). It was reported that after exercise, females exhibited a decrease in peak knee extension moment by approximately 22% (Kernozek et al., 2007). Additionally, when compared to males, females landed with lesser knee flexion and greater knee anterior shear force, but across all subjects, exercise caused subjects to land in a way that reduced the magnitude of ATSF by a mean of 29% (Kernozek et al., 2007). It was proposed that females exhibited performance and landing changes that increased the risk of non-contact ACL injury following fatigue (Kernozek et al., 2007).

While several studies have found a decrease in internal knee extension moment following exercise (Chappell, 2005; Kernozek et al., 2007; Orishimo & Kremenic, 2006), this kinetic alteration may be favorable with respect to ACL injury risk unless it is coupled with lesser knee flexion. Shimokocki et al (2013) investigated the effects of different sagittal plane body positions on lower extremity biomechanics and muscle activation during single-leg landings. They utilized both an upright landing and a lean forward landing to determine which biomechanical position would result in the most protection for the ACL (Shimokochi et al., 2013). It was reported that the upright landing posture resulted in an increase in both the peak knee extensor moment and the quadriceps muscle activation, while in turn decreasing the knee flexion angle (Shimokochi et al., 2013). Conversely, the lean forward landing decreased the internal knee extension moment, decreased quadriceps activation, and increased the knee flexion angle (Shimokochi et al., 2013). These results indicated that an upright landing posture is more harmful for the integrity of the ACL, due to the energy absorption strategy being less effective in the upright landing posture (Shimokochi et al., 2013). The results of this study support the notion that the decrease in internal knee extension moment following exercise cited by the other studies would be favorable for protecting the ACL from injury, but only if it were not accompanied by lesser knee flexion during landing.

2.4a Conclusion

It has been clearly demonstrated that after exercise, subjects tend to land with lesser knee flexion, with this movement alteration likely detrimental to the ACL. It has also been found that individuals exhibit a decrease in internal knee extension moment after exercise. By itself, this change is not necessarily bad because it would likely reduce the magnitude of ACL strain resulting directly from quadriceps loading. However, this potential benefit is offset when coupled with a decreased knee flexion angle following fatigue as the quadriceps force produced is directed more anteriorly, which can increase ACL loading. Additionally, some studies have shown that subjects utilize a greater range of knee flexion motion after exercise, which could be an adaptation for the fact that they are unable to produce a large amount of quadriceps force, possibly due to muscle weakness following exercise. This adaptation can be detrimental to the ACL due to the decreased knee flexion angle that occurs at landing. This erect landing as a suboptimal position for the ACL is supported by the decreased KEM and lesser knee flexion reported in several studies as well as investigations that have highlighted the harmful impact of a more erect posture on the ACL.

2.5 Energy Absorption

2.5a Definition and Quantification

As the body contacts the ground during landing, eccentric contraction of the extensor muscles of the hip, knee, and ankle reduce the body's center of mass velocity and absorb the kinetic energy of the system (Norcross, Blackburn, Goerger, & Padua, 2010). These eccentric muscle contractions serve as the major mechanism for energy absorption during human movement (Winter, 2005). An eccentric contraction, in which the muscle is characterized as doing negative work, occurs when the net internal joint moment (M) and joint angular velocity (ω) act in opposite directions (Winter, 2005). Conversely, a concentric contraction, or positive joint work, is when the net internal joint moment and joint angular velocity act in the same direction (Winter, 2005). As a result, the magnitude of energy absorption (negative joint work) resulting primarily from eccentric contraction is quantified by integrating the negative portion of the joint power curve ($P = M \times \omega$).

2.5b Role of EA During Movement

Energy absorption takes place during movement in order for the body to slow down its center of mass (Winter, 2005); or more simply put, energy absorption is utilized so that we do not fall down. In some instances of movement when there is not an external load present, such as during the swing phase of walking or running, both the energy generation and absorption are merely required to move the limbs (Winter, 2005). The energy generation will occur through concentric contractions of the individual muscles in order to move the body (Winter, 2005), while energy is actively absorbed in the muscles through eccentric contractions and distributed over multiple body segments (Winter, 2005). In addition to active absorption by the muscles, there is also passive absorption in the bones, ligaments, and articular cartilage (Coventry et al., 2006), which can raise reasons for concern. If the force demand placed on the passive structures exceeds what they are able to withstand, the integrity of these structures could be compromised. Devita and Skelly (1992) proposed that if greater absorption took place in the muscles, this might reduce the strain placed on the passive structures. Similarly, Voloshin et al (1998) proposed that if there is greater energy dispersion across the active mechanisms of absorption, there is potential to decrease the amount of strain placed on the passive structures, reducing the overall risk of injury. As a result, greater energy absorption by the muscle-tendon unit is generally considered favorable with respect to injury risk (Devita & Skelly, 1992). However, recent work indicates that it is important to consider not only the magnitude, but the timing, of when during landing this absorption is taking place (Norcross et al., 2010).

Norcross et al (2010) investigated the lower extremity energy absorption during double leg jump landings over three intervals: 1) the initial (INI) impact phase over the 100 ms immediately following initial ground contact, 2) the total (TOT) landing phase, from initial ground contact to the minimum vertical position of the whole body center of mass (COM Min), and 3) the terminal (TER) phase of landing, which was the time following the first 100 ms after ground contact until COM Min.. This study reported that during the initial phase of landing (the first 100 ms immediately following ground contact), the magnitude of knee energy absorption was significantly associated with lesser peak knee flexion (Norcross et al., 2010). At the hip, greater energy absorption during the initial phase of landing was related to greater anterior tibial shear force at the knee (Norcross et al., 2010). In contrast, during the TER phase, greater energy absorption at the knee was significantly associated with greater peak knee flexion angles, (Norcross et al., 2010), suggesting that greater absorption throughout landing may be a more protective strategy for the ACL.

Additionally, Norcross et al (2013) investigated the biomechanics of individuals classified as using low-, moderate-, or high-sagittal-plane initial (INI) energy absorption. The initial impact phase was defined as the first 100 ms immediately following initial ground contact (Norcross et al., 2013). Subjects were instructed to perform a double-leg jump landing and transition immediately into a vertical jump (Norcross et al., 2013). The primary finding of the study was that individuals who absorbed a greater magnitude of energy during the INI landing period possessed a movement strategy that could result in greater loading on the ACL (Norcross et al., 2013). Specifically, it was reported that greater knee extension moment and greater anterior tibial shear force resulted from greater sagittal plane energy absorption in the first 100 ms immediately after ground contact (Norcross et al., 2013). These biomechanical factors are likely indicative of greater quadriceps force, which can potentially induce greater ACL loading (Norcross et al., 2013).

It has been established that both internal knee extension moment and anterior tibial shear force are known to be biomechanical factors related to ACL injury (Chappell, 2005; Kernozek et al., 2007; Orishimo & Kremenic, 2006). ACL injuries commonly occur within the first 100 ms of landing (Cerulli, Benoit, Lamontagne, Caraffa, & Liti, 2003; Koga et al., 2010), so it is problematic that both factors have been reported to occur within that timeframe. Additionally, if the active absorbers are generating a greater amount of force with the knee in a vulnerable position, this could cause ligamentous failure. An altered distribution of energy absorption increases known biomechanical risks during the initial 100 ms of landing, and if the strain shifts from the active to the passive structures, it could compromise the integrity of the ligament and increase the amount of injuries.

2.5c Conclusion

Energy absorption is quantitatively determined by integrating the negative portion of the joint power curve (product of the internal joint moment and joint angular velocity) during a movement. Eccentric contractions, or negative joint work, occur in the muscles in order to decrease the center of mass velocity and stop the body from hitting the ground. Energy is absorbed during demanding tasks primarily in the muscles, but there is also passive absorption in the bones, ligaments, and articular cartilage. The absorbed energy is distributed over several areas, and when the active absorbers such as the quadriceps induce strain on specific structures as they eccentrically contract, they can cause a resultant anterior tibial shear force that stresses the ACL. This is especially apparent during the first 100 ms of landing, as it has been shown that greater energy absorption during this time period is significantly associated with greater knee extension moment and anterior tibial shear force.

2.6 Energy Absorption and Exercise

Lower extremity absorption occurs through eccentric contractions at the hip, knee, and ankle (Winter, 2005). Several studies have shown that females tend to function with generally less knee flexion, resulting in a more erect or stiff posture at the knee joint (Chappell, Creighton, Giuliani, Yu, & Garrett, 2006; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Koga et al., 2010; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001). Decreased knee flexion has also been exhibited extensively following exercise (Chappell, 2005; Cortes et al., 2013, 2012; Kernozek et al., 2007; Orishimo & Kremenic, 2006). In an erect landing posture, it has been proposed that contribution from the larger extensors (quadriceps) is minimized (Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007), which causes more passive force to be placed on the ligaments and tissues in the knee. In addition, the eccentric contraction occurring in the quadriceps increases the anterior pull on the tibial tuberosity (DeMorat, 2004).

With decreased dynamic control and increased anterior force, the ACL is largely at risk for injury.

This risk is demonstrated by Norcross et al (2010), which reported that during a jump landing, lower extremity energy absorption is directly related to the biomechanical factors that are associated with ACL injury. It was concluded that in the initial phase of landing (100 ms after the point of initial ground contact), greater peak anterior tibial shear force was significantly associated with greater energy absorption at the ankle (Norcross et al., 2010), which was found by Decker et al (2003) to be associated with a more erect landing posture, or lesser knee flexion. As stated, a more erect landing posture is dangerous due to the increased ligamentous injury risk from the anterior pull of the quadriceps and the increased ACL elevation angle. This posture is especially dangerous in the first 100 ms after ground contact, since most ACL injuries occur during this time period (Cerulli et al., 2003; Koga et al., 2010).

The addition of fatigue or exercise in relation to energy absorption should be considered in regards to its contribution to the safety of the ACL. It has been suggested that fatigue could alter the distribution of absorption, which would in turn increase the risk of injury (Coventry et al., 2006; Voloshin et al., 1998). Coventry et al (2006) utilized a fatigue landing protocol consisting of a single-leg drop landing from a trapeze bar, a single-leg countermovement jump, and five bodyweight single-leg squats. Coventry et al (2006) discovered that a smaller range of motion was present at both the hip and knee after exercise, and it is suggested that this decrease in range of motion could indicate a stiffer landing strategy. This broad finding could be attributed to the lack of difficulty and functionality of the fatigue protocol. This study also revealed that the landing strategy changed after fatigue. However, the same level of shock attenuation, or energy absorption, was maintained (Coventry et al., 2006). This result raises the question as to how the same energy absorption was achieved with two different landing strategies.

More recently, Norcross et al (2014) investigated the influences of landing posture on energy absorption. The study investigated both flexed and erect landing postures at the knee (Norcross et al., 2014). It was reported that during the flexed condition, peak and initial contact joint angles were significantly greater when compared to the erect condition; however, these changes were not due to fatigue, but rather were controlled experimentally (Norcross et al., 2014). Interestingly, subjects absorbed the same magnitude of energy at the knee during both flexed and erect conditions; however, the way in which this was achieved was through different underlying mechanisms (Norcross et al., 2014). As mentioned, the magnitude of energy absorption is quantified by integrating the negative portion of the power curve, which consists of a combination of the knee extension moment and angular velocity (Winter, 2005). During the flexed condition, subjects generated greater mean knee extensor moment, but had lesser mean knee angular velocity when compared to the erect condition (Norcross et al., 2014). From this result, Norcross et al (2014) proposed a hypothesis to address the lack of change in energy absorption, in which the magnitude of energy absorption at the knee might be maintained when landing with lesser knee flexion, by reducing the mean knee extensor moment and increasing the knee flexion angular velocity, which could result in the same energy absorption (Norcross et al., 2014). The researchers suggest that when knee extensor moment production is compromised, such as after fatigue, an individual could implement a more erect landing posture in order to still complete the task at hand. However, this adaptation could place the knee at an increased injury risk with respect to ACL injury (Norcross et al., 2014).

This idea proposes the question of *how*? How do individuals change their landing strategy, but retain the same amount of energy absorption? One potential reason for a more erect landing posture could be to decrease the amount of force required by the quadriceps by positioning the ground reaction force closer to the knee joint center. This would effectively decrease the internal extension moment requirement, but allow for increased knee joint angular velocity, such that the total magnitude of EA is maintained. However, the decreased amount of knee flexion at initial contact could place more strain on the ACL (DeMorat, 2004). The decreased amount of knee flexion causes a greater proportion of the quadriceps force to be directed anteriorly and increases the ACL elevation angle, which reduces the ligaments ability to withstand stress (Yu & Garrett, 2007).

2.6a Conclusion

Several studies have looked at energy absorption across the lower extremity joints. Collectively, their findings have concluded that increased absorption at the ankle is associated with a more erect landing posture. It has been clearly demonstrated that after fatigue, landing with decreased knee flexion and a large amount of anterior tibial shear force can be detrimental to the integrity of the ACL; however, it is not known why this landing posture occurs. Studies have shown that fatigue can alter the distribution of energy absorption, which could increase the risk for injury; however, these changes have been shown without fatigue as well. It has been reported that individuals landed with lesser peak and initial contact joint angles without fatigue as an intervention. Further, it was reported that the same magnitude of energy absorption was maintained, but the underlying mechanisms of energy absorption were different. One possible reason for lesser knee flexion during fatigued landings could be that individuals are attempting to decrease the internal knee extension moment required, while simultaneously increasing the joint angular velocity so that they can maintain the same amount of energy absorption. While this landing strategy would be efficient for successfully completing a given movement task, the use of lesser knee flexion could put the ACL at a significantly greater risk for injury.

2.7 Summary

ACL injuries are common, costly, and debilitating. They occur during cutting and landing maneuvers, typically in younger female athletes, when the body is trying to change direction quickly. After surgery, individuals are at an increased risk to sustain a second injury, and are also at risk for early-onset osteoarthritis later in life and eventually total knee replacement surgery. There are several risk factors related to ACL injury risk; however, this study will focus on evaluating the interplay between internal knee extension moment and knee flexion angle due to their influence on the risk of sustaining an ACL injury. In the sagittal plane, a strong eccentric quadriceps contraction at lesser knee flexion angles increases the anterior tibial shear force on the tibia and changes the type of loading (tensile vs. shear) placed on the ACL. Further, after exercise, a decrease in knee extension moment occurs. While this is not necessarily bad, this kinetic change is usually coupled with a less flexed knee position at initial contact. Lastly, it has been shown that greater knee flexion range of motion may be utilized after exercise, which could serve as an adaptation to maintain the same magnitude of knee energy absorption despite a decreased force producing capacity of the quadriceps. However, this potential adaptation could place the knee at increased risk for ACL injury. As a result, the purpose of this study is to: 1) evaluate the influence of exercise on the magnitude of knee EA during a single-leg land and cut task, 2) identify whether exercise influences the individual biomechanical determinants (internal knee extension moment and angular velocity) of knee EA, and 3) evaluate the influence of exercise on knee flexion angle at initial contact during a single-leg land and cut task.

CHAPTER 3

Materials and Methods

3.1 Subjects

Forty recreationally active females between the ages of 18-30 were recruited to participate in this study. To be eligible for inclusion, participants were required to participate in at least 150 minutes of moderate to vigorous physical activity a week (Garber et al., 2011) and to report participating in physical activity involving cutting or jumping within the previous six months. Participants were excluded if they had: 1) a prior history of ACL injury or low back, hip, knee, or ankle surgery, 2) history of any lower extremity or low back injury within the 6 months prior to data collection that limited their regular physical activity, or 3) any injuries or illnesses at the time of testing that limited their ability to perform their regular physical activity.

3.2 Subject preparation and experimental procedures

Participants reported to the Biomechanics Laboratory where they were informed of the study procedures and risks of participation before providing written consent to participate. Prior to testing, each subject was outfitted in spandex shorts and shirt and was instructed to wear their own athletic shoes for testing. Subjects completed a five-minute warm-up on a stationary bike at a self-selected speed before their height and mass of participants was recorded for biomechanical model generation. Subject's leg dominance was assessed as described by Hoffman et al (1998). In short, patients performed three tests: 1) kicking a ball, 2) stepping up onto a step, and 3) recovering after a posterior perturbation. The dominant limb was defined as the limb used at least twice to kick the ball, step up first, and utilized to regain balance.

After completion of the warm-up, several standard retro-reflective markers (27 static, 23 dynamic) were placed on the subject. Markers were placed bilaterally on the acromion process, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, anterior thigh, medial and lateral femoral epicondyles, anterior shank, medial and lateral malleoli, and on the sacrum. Markers on the foot were placed on top of the participants' shoes, with approximate locations at the calcaneus and the 1st and 5th metatarsal heads. A nine-camera motion capture system (Vicon, Inc., Centennial, CO, USA) was used to capture kinematic data during a static calibration trial and a single-leg jump-cut task.

The single-leg jump-cut task was performed as described by Frank et al (2011). Participants stood at a distance equal to 50% of their height away from the edge of a force plate (Type 4060-08, Bertec Corporation, Columbus, OH, USA), and a hurdle 17cm high was placed at a distance equal to 25% of their height in front of the force plate. They were instructed to jump off both legs over the hurdle, land with their dominant foot positioned in the center of the force plate, and cut at a 45 degree angle as quickly as possible in the direction opposite of the dominant leg (e.g., right leg dominant participants will cut toward the left). Subjects completed at least 3 practice trials and 5 successful testing trials with 30 seconds of rest between trials to minimize the potential effects of fatigue. Trials were judged successful if the participant jumped from the ground over the hurdle using both legs, landed with their entire foot on the force plate, and successfully completed the cut in the assigned direction.

After completion of the landing trials, the reflective markers were removed and the exercise protocol was explained to the participant. Subjects performed an overall exercise bout of 30 minutes, consisting of 5 cycles of exercise with each one lasting 6 minutes. Each exercise cycle began with 5 minutes of treadmill walking at a self selected pace, between 3.0 - 3.5 mph, and concluded with 1 minute of continuous jumping exercises (30 seconds of double-leg tuck jumps and 30 seconds of single-leg alternating lateral jumps). The chosen treadmill walking speed was maintained throughout the duration of the protocol. For the double-leg tuck jumps, subjects were instructed to stand in a comfortable upright position with their feet shoulder-width apart. They were told to jump explosively, and were allowed to use both arms to assist in bringing their knees as close to their chest as possible for the tuck position. On return to the starting position, they repeated as many tuck jumps as possible for 30 seconds. The single-leg lateral jumps also began in a comfortable upright position. Subjects were instructed to explosively jump from one leg to the other in a lateral direction for 30 seconds. After the jumping bout was completed, they returned to the treadmill. The treadmill started at 0% of incline, and after each minute of exercise, the treadmill incline increased by 1% until the maximal incline (15%) was reached. Borg rated perceived exertion (RPE) scale, which ranges from 6 to 20, was reported before the beginning of the exercise session and at the end of the each 6-minute session. The exercise protocol is summarized in Table 1.

At the conclusion of the exercise bout, the reflective markers were replaced using the same placement as in the pre-exercise condition and a new five-second static calibration trial was recorded. The directions for the single-leg land and cut task were explained again before subjects completed 5 more successful testing trials.

3.3 Data sampling and reduction

Vicon Motus motion capture software (Vicon, Inc., Centennial, OH, USA) was utilized to capture kinematic and force plate data, which was sampled at 120 Hz and 1560 Hz, respectively. The MotionMonitor motion analysis software was used for biomechanical model generation (Innovative Sports Training, Chicago, IL, USA) following import of the raw three-dimensional kinematic coordinates and kinetic data. The ankle joint center was defined as the midpoint of the medial and lateral malleolus, the knee joint center defined as the midpoint of the medial and lateral femoral epicondyle markers, and the hip joint center defined using external landmarks as described by Bell et al (1989). The local coordinate system 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. Kinematic and force plate data was lowpass filtered at 10 Hz (4^{th} order zero-phase lag Butterworth) with kinematic data time-synchronized to kinetic data and re-sampled 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 was defined about the ankle as the foot relative to the shank, about the knee as the shank relative to the thigh, and about the hip as the thigh relative to the sacrum. Instantaneous joint angular velocities were calculated as the 1st derivative of angular position.

Net internal joint moments of force at the ankle, 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).

Custom computer software (LabVIEW, National Instruments, Austin, TX) was utilized to calculate knee EA, mean internal KEM, and mean knee flexion angular velocity during the 100 ms immediately following initial contact (vertical ground reaction force > 10 N) for all single-leg jump-cut trials. The knee joint power curve was determined by multiplying the knee joint angular velocity and net internal knee moment ($P = M \times \omega$). Negative mechanical joint work was calculated by integrating the negative portion of the joint power curve during the initial 100 ms of landing (Decker et al., 2003; DeVita, Janshen, Rider, Solnik, & Hortobágyi, 2008; Schmitz et al., 2007), with negative joint work values representing energy absorption by the muscle-tendon unit (Winter, 2005). In order to simplify interpretation during data analysis, these values were assigned positive by convention. The same custom software was also used to identify knee flexion angle at initial contact and the mean internal knee joint moment and mean knee flexion angular velocity during the initial 100 ms of landing. Mean values for all variables were calculated across the five single-leg jump-cut trials in each condition (pre vs. post-exercise) for each subject.

3.4 Statistical Analyses

Knee flexion angle at initial contact, knee EA, mean KEM, and mean knee flexion angular velocity during the pre-exercise and post-exercise sessions were compared using individual dependent samples *t*-tests. All analyses were conducted using commercially available software (SPSS 23.0, SPSS Inc., Chicago, IL, USA) with statistical significance established *a priori* as $\alpha \leq 0.05$. The statistical analysis plan is summarized in Table 2.

Table 1. Description of Exercise Protocol by Session

Set	Time	Activity	Incline
1	Begin (0 minutes) – 6 minutes	Treadmill x 5min Tuck Jumps x 30 sec Lateral Jumps x 30 sec	0% - 5%
2	6 minutes – 12 minutes	Treadmill x 5min Tuck Jumps x 30 sec Lateral Jumps x 30 sec	5% - 10%
3	12 minutes – 18 minutes	Treadmill x 5min Tuck Jumps x 30 sec Lateral Jumps x 30 sec	10% - 15%
4	18 minutes – 24 minutes	Treadmill x 5min Tuck Jumps x 30 sec Lateral Jumps x 30 sec	15%
5	24 minutes – Finish (30 minutes)	Treadmill x 5min Tuck Jumps x 30 sec Lateral Jumps x 30 sec	15%

Table 2. Statistical Analysis Plan

Research Questions	**Independent Variables**	**Dependent Variable(s)**	**Statistical Test**
What is the influence of exercise on the magnitude of lower extremity energy absorption at the knee during a single-leg land and cut task?	1) Exercise - Pre-exercise - Post-exercise	1) Knee Energy Absorption	Dependent Samples *t*-test
What is the influence of exercise on the individual biomechanical determinants (internal knee extension moment and joint angular velocity) on energy absorption?	1) Exercise - Pre-exercise - Post-exercise	1) Mean Internal Knee Extension Moment 2) Mean Joint Angular Velocity	Separate, Dependent Samples *t*-test
What is the influence of exercise on knee flexion angle at initial contact during a single-leg land and cut task?	1) Exercise - Pre-exercise - Post-exercise	1) Knee Flexion Angle at Initial Contact	Dependent Samples *t*-test

CHAPTER 4

Results

Data was collected for 40 recreationally active female volunteers (age = 21.0 ± 1.7 years; height = 167.4 cm ± 7.81 cm; mass 65.89 kg ± 8.54 kg). However, a total of 7 subjects were eliminated from final analysis due to their inability to complete the exercise protocol during the testing session and 1 subject was removed due to errors during data collection. As a result, the final sample analyzed contained 32 recreationally active females (age = 21.1 ± 1.7 years; height = 169.1 cm ± 6.23 cm; mass 65.53 kg ± 8.01 kg).

With respect to knee flexion angle at initial contact, subjects landed no differently in the post-exercise condition when compared to the pre-exercise condition (t_{31} = -1.416, p = .167) (Figure 1). Post-exercise, subjects absorbed significantly lesser energy at the knee when compared to their pre-exercise trials (t_{31} = -2.586, p = .008) (Figure 2). Similarly, subjects exhibited significantly lesser internal knee extension moment post-exercise (t_{31} = -3.098, p = .004) (Figure 3). Finally, subjects exhibited significantly lesser mean joint angular velocity at the knee in the post-exercise condition (t_{31} = 4.016, p < .001) (Figure 4). The results are summarized in Table 3 below.

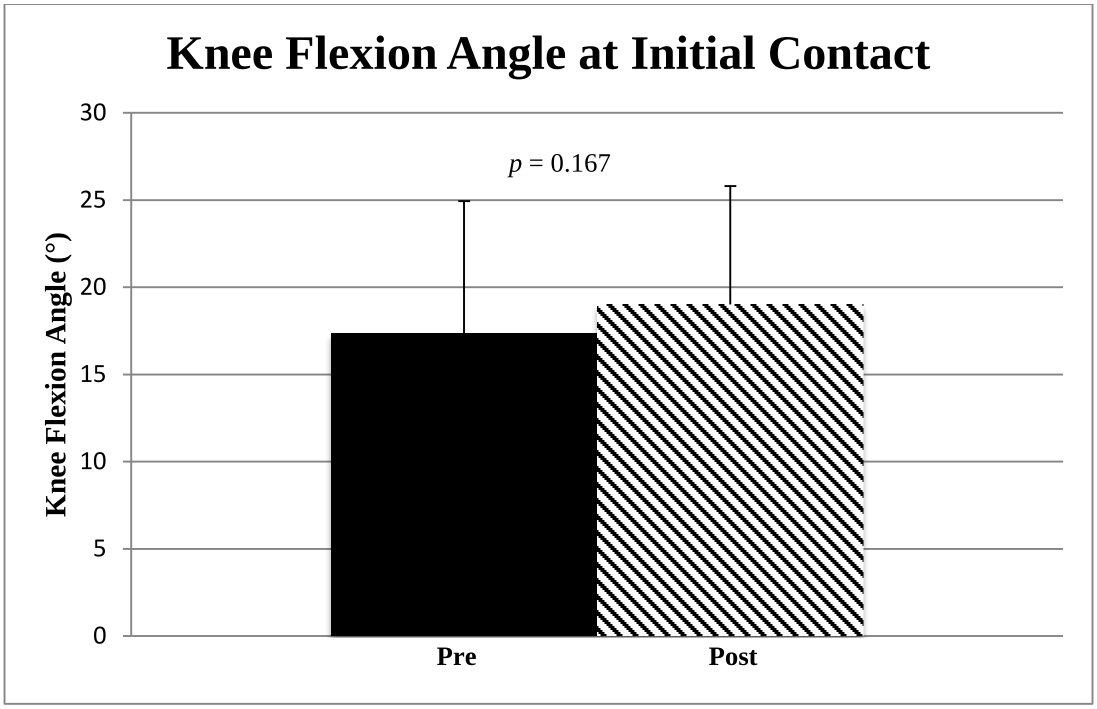

Figure 1. Knee Flexion Angle at Initial Contact Pre-Exercise and Post-Exercise

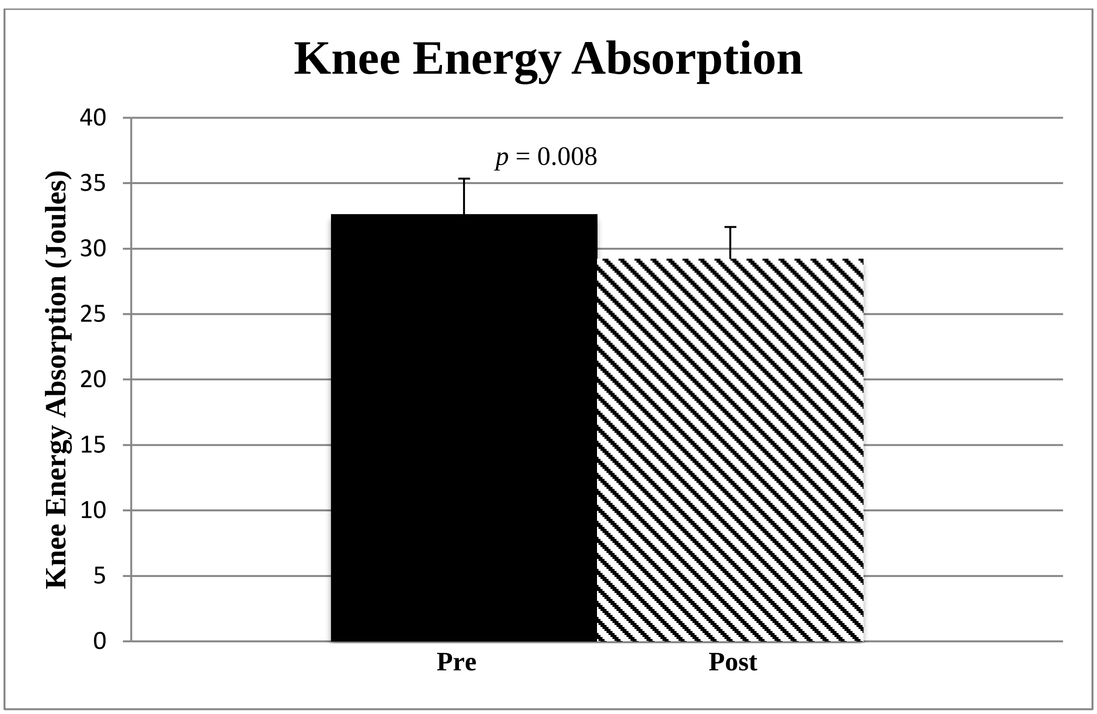

Figure 2. Knee Energy Absorption During the First 100 ms of Landing Pre-Exercise and Post-Exercise

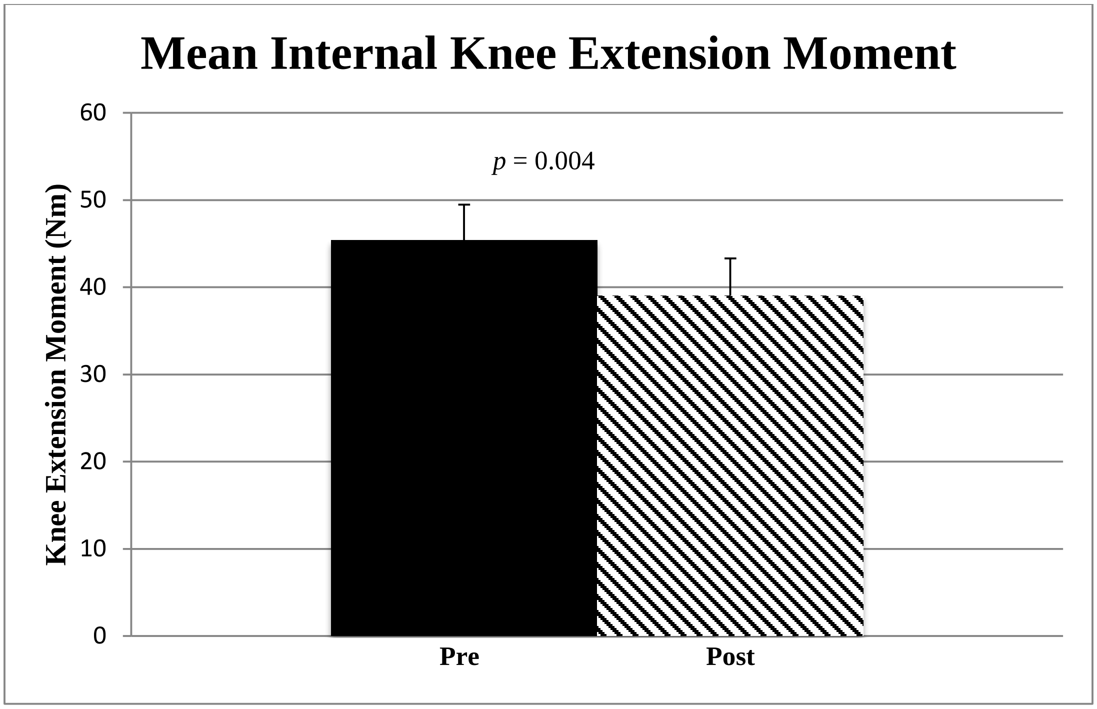

Figure 3. Mean Internal Knee Extension Moment During the First 100 ms of Landing Pre-Exercise and Post-Exercise

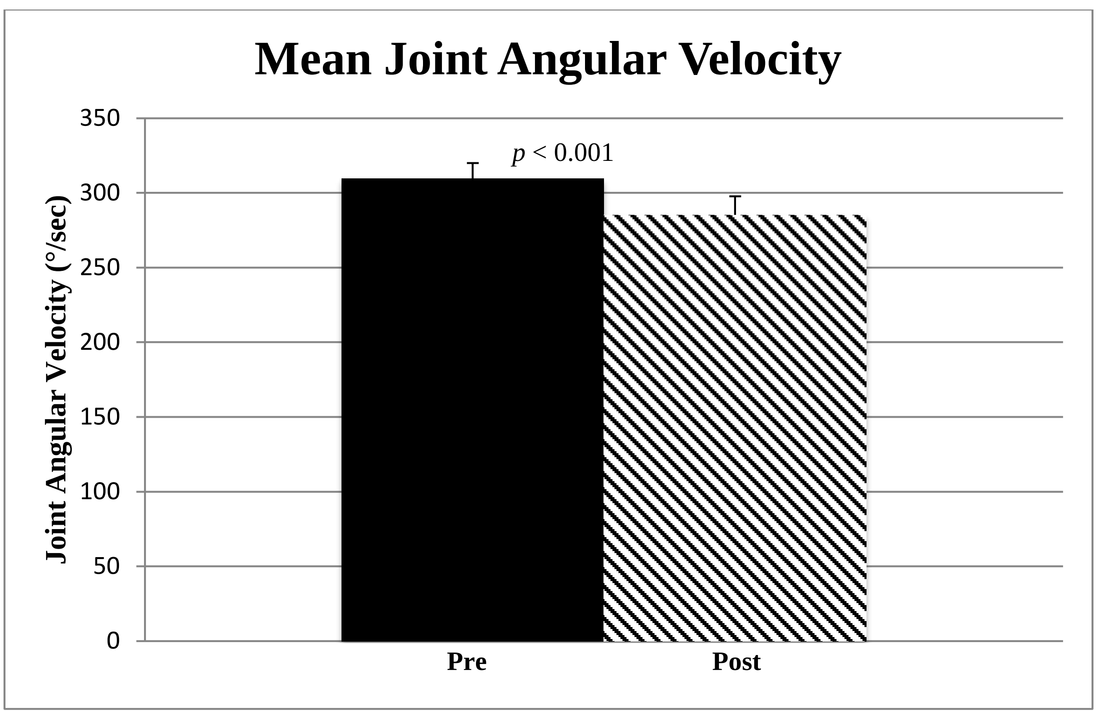

Figure 4. Mean Joint Angular Velocity During the First 100 ms of Landing Pre-Exercise and Post-Exercise

Table 3. Summary of Results

	Pre – Exercise	Post – Exercise	Mean Difference	95% Confidence Interval of the Difference
Knee Flexion Angle at Initial Contact (°)	17.35 ± 7.59	19.01 ± 6.80	-1.66	(-4.04, 0.73)
Knee Energy Absorption (J)	-32.60 ± 15.48	-29.20 ± 13.83	-3.40	(-5.82, -0.97)
Internal Knee Extension Moment (Nm)	-45.38 ± 23.03	-39.05 ± 24.05	-6.33	(-10.50, -2.16)
Joint Angular Velocity (°/sec)	309.59 ± 58.34	285.32 ± 70.34	24.27	(11.95, 36.60)

CHAPTER 5

Discussion

The primary finding of this investigation is that female participants landed with significantly lesser knee energy absorption during the 100 ms immediately after ground contact following the completion of a standardized exercise protocol despite using the same initial contact knee flexion position. The lesser EA following exercise was the result of a 14% reduction in mean internal knee extension moment coupled with an 8% reduction in mean knee flexion angular velocity post-exercise.

Contrary to our hypothesis, there was no significant change in knee flexion angle at initial contact following exercise in this sample of female participants performing a single-leg jump-cut task (Figure 1). While unexpected, there have been a few previous investigations that also did not report a difference in knee flexion angle at initial contact before and after some type of exercise intervention. Kernozek et al (2007) measured the landing biomechanics of female subjects as they performed 50 cm drop landings before and after completing an exercise protocol consisting of at least 4 repetitions of unlimited squats with a 90 second rest in between sets (Kernozek et al., 2007). They found that females landed with almost the same knee flexion angle at initial contact pre- (7.78°) and post-exercise (7.86°) (Kernozek et al., 2007). Additionally, Orishimo and Kremenic (2006) identified no change in knee flexion angle at initial contact during single-leg hops in female subjects after the completion of two sets of 50 step-ups onto a 30 centimeter box (Orishimo & Kremenic, 2006). While these two studies agree with our findings, the majority of studies investigating landing biomechanics following exercise have reported that females tend to land with lesser knee flexion post-exercise compared to pre-exercise (Chappell, 2005; Cortes et al., 2013; Lucci et al., 2011). Given this, it is possible that the single-leg jump-cut used in this investigation did not impose a sufficient demand that required a change in initial contact knee position. It may be that the subjects performed the single-leg jump-cut by simply redirecting their whole body center of mass along the new direction of travel rather than by rapidly decelerating and then accelerating along the new path. As a result, the demand on the quadriceps to eccentrically contract to decelerate their center of mass may not have been sufficiently high to cause the expected change in knee flexion angle following exercise.

In support of this idea, previous work by Cortes et al (2013), Lucci et al (2011), and Chappell et al (2005) that have observed lesser knee flexion at initial contact following exercise used tasks that are relatively more demanding than the single leg jump-cut. Cortes et al (2013) and Lucci et al (2011) had subjects complete a single-leg land and cut similar to the one used in this study, but between the land and cut, subjects had to reach down to touch the force plate with their hand. Chappell et al (2005) had subjects start with a running approach, take off from a single leg, and perform a double-leg landing that transitioned immediately into a maximum vertical jump. Since subjects had to touch the force plate in the single-leg landing, they may have been forced to decelerate their center of mass more, in order to complete the task correctly. In the movement utilized by Chappell et al (2005), subjects had to decelerate quickly from a running approach in order to stop their horizontal velocity and transition immediately into the maximum vertical jump. Therefore, the single-leg jump-cut task used in this study might not have required enough deceleration in order to facilitate the expected change in knee flexion angle at initial contact. However, despite the lack of change in initial contact knee position, we present novel findings that females absorbed approximately 10% less energy at the knee during the initial 100 ms of a single-leg jump-cut task following the completion of an intense exercise bout, and that this decrease in knee EA was driven by a reduction in both mean knee extension moment and knee flexion angular velocity post-exercise.

The results for internal knee extension moment were consistent with our hypothesis as the mean moment decreased by 14% post-exercise. This kinetic change is also in agreement with previous investigations performed by Chappell et al (2005), Orishimo and Kremenic (2006), and Kernozek et al (2007) that reported 14%, 16%, and 22% reductions, respectively, in the internal knee extension moment post-exercise. However, contrary to our hypothesis, we also found that the mean knee flexion angular velocity during the initial 100 ms of landing decreased by approximately 8% (25°/s) post-exercise. Consequently, the magnitude of knee EA following exercise was not maintained as the combination of lesser knee extension moment and lesser knee flexion angular velocity resulted in the 10% reduction in knee energy absorption following exercise.

In regards to energy absorption during landing, it has been reported that the total magnitude of lower extremity energy absorption and the amount of energy absorbed by each joint can be influenced by the “stiffness” of the landing (Zhang, Bates, & Dufek, 2000). Specifically, “stiff” landings are generally characterized by lesser joint flexion and result in a greater proportion of energy absorption at the ankle and a lesser proportion of energy absorption at the knee compared to softer landings (DeVita et al., 2008; Schmitz et al., 2007; Zhang et al., 2000). Therefore, it is possible that the knee EA changes that we observed were due to female subjects adopting a stiffer landing strategy post-exercise. To investigate this notion, we performed secondary analyses in which we identified that the knee contribution to total lower extremity EA during the initial 100 ms of landing fell from 40.8% pre-exercise to 39.0% post-exercise, while the ankle contribution increased from 53.3% pre-exercise to 55.5% post-exercise. While these changes in relative joint contributions to EA are not substantial, it is important to note that the greater contribution of the ankle compared to the knee before and after exercise suggests that females in this investigation performed relatively "stiff" landings both before and after the exercise protocol. However, following exercise, females had 3.4° less knee flexion displacement (Pre-exercise = 39.8 ± 8.1°, Post-exercise = 36.4 ± 8.8°, $p < .001$), a 15 ms shorter stance time (Pre-exercise = 375.0 ± 48.2, Post-exercise = 359.3 ± 39.5, $p = 0.011$), and 9% less total lower extremity EA (Pre-exercise = 121.0 ± 24.3, Post-exercise = 112.45 ± 27.7, $p = 0.001$) compared to before exercise which suggests that the overall landing "stiffness" was increased following exercise. Devita and Skelly (1992) and Voloshin et al (1998) have both proposed that greater energy absorption by the active musculotendinous structures might reduce the strain placed on passive structures such as ligament and bone. As a result, the use of a stiffer landing strategy following exercise by females in the current investigation could result in an increased risk for lower extremity injury.

Previous studies investigating EA have tended to use some variation of a double-leg or single-leg landing task (Decker et al., 2003; Kulas, Schmitz, Shultz, Watson, & Perrin, 2006; Schmitz et al., 2007; Yeow, Lee, & Goh, 2009). However, the use of different landing tasks has resulted in differing results in regards to individual joint contributions to lower extremity EA (Decker et al., 2003; Kulas et al., 2006; Schmitz et al., 2007; Yeow et al., 2009). During a double-leg landing from 60 centimeters, the knee has been reported to absorb most of the energy – ranging from 47% to 61% (Decker et al., 2003; Kulas et al., 2006). However, during a barefoot double-leg landing from a shorter height of 30 centimeters, the hip absorbed the most energy at 52.8% (Yeow et al., 2009). Finally, during a single-leg terminal landing task, the ankle has been reported to absorb 88% of the total energy (Schmitz et al., 2007). While we chose to utilize a single leg jump-cut task to mimic one reported mechanism of ACL injury, subjects in this investigation absorbed the majority of energy at the ankle (54.4%) and not at the knee (39.9%) across both conditions, which was unexpected. As a result, it is not known if the use of a more stiff landing posture following exercise would be consistent for other types of landing tasks.

Finally, given the lack of change in knee flexion angle at initial contact following exercise, it is possible that the exercise protocol that was used was not demanding enough to cause the more extended knee position that was expected to occur following exercise based upon previous studies (Chappell, 2005; Cortes et al., 2013; Kernozek et al., 2007; Mclean et al., 2007; Schmitz et al., 2014). However, we do not believe that this is the case for several reasons. First, the exercise protocol for this study was adapted directly from previous studies that have shown a decrease in neuromuscular activation, maximum voluntary isometric contractions (MVIC), or knee extension torque following the completion of the protocol (Chang, Kim, Hertel, & Hart, 2014; Kuenze, Hertel, & Hart, 2013; Kuenze, Hertel, & Hart, 2014; Stern, Kuenze, Herman, Sauer, & Hart, 2012). Chang et al (2014) also reported that there were still quadriceps neuromuscular deficits in healthy subjects as long as 24-hours after the exercise was completed. With respect to the demands of the protocol, 7 subjects (17.5% of the total participants tested) were not able to complete the exercise protocol without reducing the incline of the treadmill, despite performing an average of 150 minutes per week of moderate to vigorous activity. Further, the average ending Borg rating of perceived exertion across all subjects was 18, which is between "very hard" and "very, very hard" on the scale, indicating that participants subjectively perceived being physically tired. It could be that while the protocol was tiring, it did not incorporate enough explosive movements for a long enough period of time. Other studies have used several variations of squat jumps, step-ups, sprints, and agility ladders for the majority of the protocol (Chappell, 2005; Cortes et al., 2013; Lucci et al., 2011). However in the current study, an explosive task was performed, but only for two-and-a-half minutes in total. It is also possible that the magnitude of the exercise effect was not large enough to be detected after completion of the exercise protocol. In the current study, the time from the end of exercise to the beginning of post-testing was approximately 15-20 minutes, which could have been enough time for the effect of the exercise protocol to have "worn off". Finally, as noted previously, it is also a possibility that the single-leg jump-cut task that we had participants perform was not difficult enough to cause participants to modify their landing mechanics. The results of Norcross et al (2010) and Decker et al (2003) indicate that 2-3 times the amount of energy absorption from the knee was required to complete a double leg jump landing task than was required to complete the single leg jump-cut task used in the current study. It is possible that the use of a more demanding task that increased the demand for eccentric quadriceps control may have altered our results.

CHAPTER 6

Conclusion

This investigation provides novel findings that, following exercise, physically active females absorbed significantly less energy at the knee during the 100 ms immediately following ground contact despite having the knee positioned in a similar amount of flexion at initial contact. The 10% reduction in knee EA following exercise was driven by the combination of lesser knee extension moment and lesser knee flexion angular velocity which suggests that females adopted a stiffer landing strategy. While the magnitude of the EA reduction observed during the single-leg jump-cut in this investigation is probably not clinically meaningful, it is likely that the use of a stiffer landing strategy following exercise during more demanding movement tasks might result in increased loading of passive structures and greater lower extremity injury risk. However, the magnitude and individual joint contributions to lower extremity EA are influenced by the type of movement task performed. Therefore, future research is necessary to confirm these results in other types of landing tasks.

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