AN ABSTRACT OF THE THESIS OF

Baker Cronin for the degree of Master of Science in Exercise and Sport Science presented on June 10, 2015

Title: The Influence of Hip Abduction, Extension, and External Rotation Rate of Torque Development on Frontal-Plane Biomechanics during Single-Leg Jump-Cuts

Abstract approved:

__________________________________________________________________________________

Marc F. Norcross

Anterior cruciate ligament injuries are relatively common in female athletes with greater frontal plane knee motion and loading during a landing task predictive of injury. As a result, decreasing knee abduction motion and loading is important for ACL injury prevention. While increased strength of the hip abductors, extensors, and external rotators has been commonly theorized to improve eccentric control of hip adduction and internal rotation, and thus decrease frontal plane knee motion and loading, previous work evaluating the relationship between peak torque of these muscle groups and frontal plane biomechanics is equivocal. However, as the time to generate peak torque exceeds the time during which frontal plane motion occurs during movement tasks, the capacity to rapidly develop torque may be more closely related to frontal plane biomechanics. Therefore, the objective of this study was to examine the influence of hip abduction, extension, and external rotation rate of torque development (RTD) and frontal-plane hip and knee biomechanics. Forty recreationally active females performed maximal isometric contractions and single-leg jump-cuts. From recorded torque, motion capture, and ground reaction force data, hip RTD and frontal plane hip and knee
biomechanics were calculated. For each RTD measure, jump-cut biomechanics were compared between participants in the highest (High) and lowest (Low) tertiles. No differences in frontal plane biomechanics were identified between High and Low hip abduction RTD groups. However, those in the High hip extension or High external rotation RTD groups had lesser hip adduction and knee abduction displacements. The results suggest that in movements such as cutting in which the hip is abducted and flexed, the ability of the gluteus medius to control frontal plane motion is likely reduced due to the position of the hip, but that the upper portion of the gluteus maximus might control frontal plane hip motion by functioning as a hip abductor. The results highlight an important role the gluteus maximus may play in controlling frontal plane hip and knee motion during cutting tasks.
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The Influence of Hip Abduction, Extension, and External Rotation Rate of Torque Development on Frontal-Plane Biomechanics during Single-Leg Jump-Cuts

by

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A THESIS

submitted to

Oregon State University

in partial fulfillment of the requirements for the degree of

Master of Science

Presented June 10, 2015
Commencement June 2015
Master of Science thesis of Baker Cronin presented on June 10, 2015

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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.

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Baker Cronin, Author
ACKNOWLEDGEMENTS

First off, I would like to offer my sincerest gratitude to Dr. Marc Norcross for his guidance, patience, and wisdom. He has been so instrumental in helping me design this study, teaching me sound research methods, and improving my understanding of the field. He has helped improve my writing skills considerably and has always expected the most from me, which has pushed me to continue to be better. Most of all, he has improved my skills as a clinician by making everything we have worked on applicable to my burgeoning career as an athletic trainer. I have been very fortunate to work under his guidance.

I would also like to wholeheartedly thank the rest of my committee, Dr. Sam Johnson, Dr. Christine Pollard, and Dr. Viktor Bovbjerg. They have all offered excellent critiques and input as I have shaped my question and looked for understanding in my results. Dr. Johnson, in particular, was especially insightful as I worked on developing a specific question from my wide research interests. Without his help, I would not have been able to find this topic.

Finally, I would like to thank my family. They have gone above and beyond to offer me the support and guidance I have needed throughout this entire journey. They were always there as inspiration to continue to push myself even when I was consumed with doubt. I could not ask for better people to help me through this process.
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Chapter 1: Introduction

1.1 Introduction

Landing on a single leg is a common part of many sports, such as basketball, football, volleyball, soccer, and handball. With this aspect of sport comes a potential for serious injury. Of the possible injuries, anterior cruciate ligament (ACL) injuries are one of the most publicized and result in significant long term consequences (Mather et al. 2013, Wright et al 2008). ACL injury plays a significant role in athletics, with approximately 5000 injury events from 1988 to 2004 in collegiate athletics alone (Hootman et al 2007).

In the United States population, there have been more than 125,000 ACL reconstructions (ACLR) performed each year (Mall et al 2014). Following ACLR patients require 6-12 months of extensive rehabilitation before they can return to sport (Wright et al 2008). ACL injury also brings with it an increased risk of developing osteoarthritis (OA) of the knee (Li et al 2011, Lohmander et al 2004, Øiestad et al 2008). OA is a painful degeneration of the bone which can lead to the need for a total knee arthroplasty. The combination of these factors results in an estimated 7.6 billion dollars spent annually on medical services for conditions directly related to the initial ACL injury and other costs related to lost earnings, disability, and decreased quality of life (Mather III et al. 2013).

Sudden deceleration during landing or cutting is a commonly reported mechanism for ACL injury, and injuries sustained in this way are referred to as non-contact ACL injury events (Boden et al 2000, Olsen et al 2004). Non-contact ACL injuries typically occur during landings which combine small knee flexion angles and either frontal and/or transverse plane knee
motion with concomitant loading in those planes (McLean et al 2004, Quatman et al 2010, Shimokochi and Shultz 2008). Hewett et al (2005) found that among female subjects who suffered an ACL injury, greater peak knee abduction angle and external knee abduction moment were significant predictors of the injury event. Due to this, substantial research has been geared toward targeting factors which can influence frontal-plane knee motion and loading (Willey and Davis 2004, Hollman et al 2013, Cashman 2012).

One factor of interest is proximal control of the femur, as excessive hip motion during closed chain activities such as landing or cutting, can influence frontal plane knee biomechanics (Powers 2010). In particular, eccentric control of hip adduction and internal rotation, is generally accepted as important since greater hip adduction and internal rotation have been associated with increased knee abduction angles during athletic movements (Powers 2003 and Padua 2005). The two hip muscles most often targeted clinically to improve eccentric control of the femur are the gluteus maximus and medius. Weakness of the gluteus medius, the primary hip abductor, could potentially result in greater hip adduction motion during landing, which in turn might lead to greater frontal plane knee motion and loading (Powers 2003). Similarly, weakness of the gluteus maximus, due to its role as an external rotator, is commonly theorized to contribute to greater knee abduction via reduced eccentric control of hip internal rotation (Ireland et al 2003, Powers 2003, Neumann 2010). However, as the gluteus maximus is functionally divided into upper and lower portions with the upper portion having a primary function of hip abduction (Grimaldi et al 2009), weakness of the gluteus maximus could also contribute to greater hip adduction motion and thereby greater frontal plane knee motion and loading during movement tasks.
The majority of previous research pertaining to hip muscle function and its relationship to frontal plane knee motion has focused on peak strength (Jacobs and Mattacola 2005, Hollman et al 2009, Hollman et al 2013, Homan et al 2013, Norcross et al 2009, Stearns et al 2013, Stearns and Powers 2014, Willy and Davis 2011). However, the results of studies evaluating the relationship between peak strength and frontal plane hip and knee motion have been inconclusive. Norcross et al (2009) examined the relationship of hip abductor (i.e., gluteus medius) and external rotator (i.e., gluteus maximus) peak strength on frontal plane knee motion during a single leg step down. Researchers found no relationship between peak eccentric hip abductor strength, eccentric hip external rotator strength, or isometric hip external rotator strength and frontal plane knee motion. This is counterintuitive as those muscles should help control frontal plane knee motion during landing (Neumann 2010, Powers 2010). It was theorized that the single leg step down task may not have been demanding enough, such that even the weakest subjects had sufficient strength to control hip and knee motion and therefore did not exhibit differences in kinematics compared to stronger participants. Homan et al (2013) examined the influence of hip abductor and external rotator peak strength on knee abduction, hip adduction, and hip internal rotation angles during landing. Researchers found subjects categorized as having lesser peak strength exhibited similar frontal and transverse plane hip and knee kinematics as participants with greater peak strength, demonstrating that peak strength alone is not necessarily a predictor of hip and knee kinematics during dynamic tasks. While this study employed a more difficult 30cm double leg jump landing task, the task still may not have been difficult enough to cause significant demands on the knee and hip in a healthy, recreational adult.
Stearns and Powers (2014) examined the effects of completing a hip strengthening protocol on knee kinematics. After 4 weeks of training, they found an increase in peak strength of the hip abductors and extensors, but only a 1.2° reduction in peak knee abduction angle during a landing task that was not deemed statistically significant ($p = 0.07$). However, the task completed in this study was once again a double leg landing followed by a jump and it may not have been challenging enough to elicit large magnitudes of knee abduction motion in otherwise healthy females. Jacobs and Mattacola (2005) did identify a significant relationship between greater peak eccentric hip abductor torque and lesser peak knee abduction angle in females completing a single-leg hopping task. However, a cross-sectional study by this same group reported that while females exhibited greater knee abduction displacement during single-leg landing and lesser eccentric hip abductor peak torque compared to males; the relationship between hip adduction or knee abduction displacements and peak hip abduction torque was not significant in either males or females (Jacobs et al 2007).

Hollman et al (2013) examined the effects of hip extensor strength and gluteus maximus recruitment on frontal plane knee kinematics during a jump-landing task. Researchers found that lesser hip extensor strength and gluteus maximus recruitment were significant predictors of greater peak knee valgus angles at the point of maximum knee flexion. While these results suggest a relationship between peak hip extensor (i.e., gluteus maximus) strength and frontal plane knee kinematics, this study did not quantify the magnitude of hip adduction and internal rotation motion (peak angels or displacements) and therefore it is not known if greater eccentric control of these motions was driving the lesser knee abduction angle that was observed. This study also did not measure hip abductor strength making it unclear what role
gluteus maximus strength may have played in contributing to lesser knee abduction during landing.

The aforementioned studies highlight the inconsistency of findings surrounding the relationships between peak hip strength and frontal-plane hip and knee biomechanics. While part of this discrepancy could be due to the different tasks used for each study, another potential limitation of the previous work is that these studies have quantified strength by measuring the peak torque produced during a maximal voluntary contraction (MVIC). However, peak torque is usually not achieved until 250-400 milliseconds (ms) (Aagaard 2003, Aagaard and Andersen 1998) after torque onset, with torque produced eccentrically taking longer to reach peak compared to concentrically generated torque (Aagaard and Andersen 1998). In contrast, ACL injuries are thought to occur within the first 60 ms of landing (Kernozek and Ragan 2008, Koga et al 2010) and maximum hip adduction during landing is usually reached within 130-150 ms after initial contact (Lephart et al 2002). Thus, rather than peak torque producing capacity, being able to develop sufficient torque of the hip musculature over a very short period of time may be more influential for controlling frontal plane hip and knee motions during landing.

One method for assessing explosive torque production is to quantify the rate of torque development (RTD) by calculating the slope of the torque-time curve during a specified time interval (Aagaard et al 2002). Given that previous research examining hip strength and frontal plane hip and knee motion has assessed peak torque with equivocal results, the ability to produce torque rapidly over a critical time period may be more closely related to control of frontal plane hip and knee biomechanics during landing than to peak torque. However, it is
crucial to measure RTD over a time frame which is relevant to frontal plane hip and knee biomechanics during landing. A 0-200 ms time frame represents an early contraction phase that provides information specific to the explosive force producing capacity of muscle function (Aagaard et al 2002). Further, this time interval is longer than the time required to reach peak hip adduction and knee abduction angles during landing, making it an appropriate interval over which to measure differences in explosive torque production that could influence frontal plane hip and knee biomechanics.

As demonstrated above, the relationship between strength and frontal plane hip and knee mechanics during landing has been examined, but with conflicting results. While this inconsistency could be due to the use of different landing tasks across studies, it is also possible that peak torque is not the most appropriate strength measure to associate with landing biomechanics because it is not representative of the rapid manner in which torque must be produced during dynamic tasks. Research evaluating the relationships between rate of torque development and frontal plane motion during functional tasks, such as single-leg jump-cuts is lacking. Since RTD can be increased with training (Holtermann et al 2007), this offers a potentially new avenue for interventions to prevent measures towards knee lower extremity injuries, such as, ACL sprains.

Therefore, the purpose of this study was to determine the influence of hip abductor, extensor, and external rotator RTD on frontal plane hip and knee biomechanics during a single-leg jump-cut task.

1.2 Research Question:

RQ 1: Is there a significant difference between individuals classified with High and Low rate of
torque development of the hip abductors and the following variables during a single-leg jump-cut task?

- Hip adduction at initial contact
- Peak hip adduction
- Hip adduction displacement
- Knee abduction at initial contact
- Peak knee abduction
- Knee abduction displacement
- Peak internal knee varus moment

**RH 1:** I hypothesized that, compared to the Low hip abductor RTD group, individuals in the High RTD group would exhibit significantly less:

- Hip adduction at initial contact
- Peak hip adduction
- Hip adduction displacement
- Knee abduction at initial contact
- Peak knee abduction
- Knee abduction displacement
- Peak internal knee varus moment

**RQ 2:** Is there a significant difference between individuals classified with High and Low rate of torque development of the hip extensors and the following variables during a single-leg jump-cut task?
• Hip adduction at initial contact
• Peak hip adduction
• Hip adduction displacement
• Knee abduction at initial contact
• Peak knee abduction
• Knee abduction displacement
• Peak internal knee varus moment

RH 2: I hypothesized that, compared to the Low hip extension RTD group, individuals in the High RTD group would exhibit significantly less:

• Hip adduction at initial contact
• Peak hip adduction
• Hip adduction displacement
• Knee abduction at initial contact
• Peak knee abduction
• Knee abduction displacement
• Peak internal knee varus moment

RQ 3: Is there a significant difference between individuals classified with High and Low rate of torque development of the hip external rotators and the following variables during a single-leg jump-cut task?

• Hip adduction at initial contact
• Peak hip adduction
• Hip adduction displacement
• Knee abduction at initial contact
• Peak knee abduction
• Knee abduction displacement
• Peak internal knee varus moment

RH 3: I hypothesized that, compared to the Low hip external rotator RTD group, individuals in the High RTD group would exhibit significantly less:

• Hip adduction at initial contact
• Peak hip adduction
• Hip adduction displacement
• Knee abduction at initial contact
• Peak knee abduction
• Knee abduction displacement
• Peak internal knee varus moment

1.3 Operational Definitions

Initial Ground Contact (IGC): The beginning of the landing period was defined as the instant when the vertical component of the ground reaction force exceeded 10 Newtons.

Dominant limb: The limb which leg subjects used at least twice for the following tasks: 1) to kick a ball for distance, 2) to step up on 25cm box, and 3) to recover from small push from behind.

Single-leg jump-cut: Subjects stood positioned 50% of their height away from a force plate. They then jumped with both feet forward toward the plate and landed on the dominant foot, centered on a force place before immediately cutting in the direction opposite of the dominant leg at a 60 degree angle.
1.4 Assumptions & Limitations

The following assumptions were made:

1. Participants performed all testing protocols to the best of their ability and with maximum effort.
2. Participants were honest regarding their prior history with respect to the inclusion/exclusion criteria.
3. The biomechanical data collected during these experiments was reliable and valid for all participants.

1.5 Delimitations

The following delimitations were made for this thesis project:

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 had no history of lower back or lower extremity surgery,
4. All participants had no current injury or illness which limited their physical activity level.
5. All participants were physically active as defined by participation in at least 150 minutes of activity a week.
6. All participants performed a cutting or jumping task within the last 6 months.
Chapter 2: Literature Review

2.1 Introduction: Landing on a single leg is a common part of many sports, such as basketball, football, volleyball, soccer, and handball to name a few. With this aspect of sport comes a potential for serious injury. Of the possible injuries, anterior cruciate ligament (ACL) injuries are among the most publicized and catastrophic.

2.2 ACL Injury: In college athletics alone, there were 4800 ACL injuries from 1988 to 2004 (Hootman et al 2007). Hootman et al (2007) found the rate of ACL injuries in college athletics increased an average of 1.3% per year over the same time span. Overall, the majority of ACL injuries take approximately 6-12 months from injury until return to play (Wright et al 2008). In the United States as a whole, the number of ACL injuries suffered per year has also increased, resulting in more than 40,000 more reconstructions per year in 2006 compared to 1994 (Mall et al. 2014). This means, in just 13 years, the number of ACL reconstructions (ACLR) performed per year increased by about 50% (86,687 to 129,836). Mall et al (2014) also found the incidence of ACLR among women has increased from 10.36 per 100,000 person-years to 18.06 per 100,000 person-years from 1994 to 2006. This means that there are nearly twice as many ACLR per 100,000 females person-years compared to just a decade earlier.

With the rise of ACL injury has come a concomitant increase in the economic impact of this injury. The lifetime burden of ACLR in the U.S. is estimated to be 7.6 billion dollars annually, accounting for surgery, rehabilitation, and costs of altered lifestyle (Mather III et al. 2013). While this is substantial, ACLR remains the treatment of choice for many patients as it is in fact cheaper to society than conservative rehabilitation, which has been estimated to result
in an annual cost of 17.7 billion dollars (Mather III et al. 2013).

Along with its great cost, ACL injury also brings with it an increased risk of developing osteoarthritis (OA) of the knee (Li et al 2011, Lohmander et al 2004, Øiestad et al 2008). OA is a painful degeneration of the bone which can lead to the need for a total knee arthroplasty. According to Øiestad et al (2008), a patient who suffers an ACL injury without injury to the meniscus has a 0-13% chance of developing knee OA more than 10 years after the injury; but if the meniscus is involved the likelihood increases to between 21% and 48%. Li et al (2011) reported similar results with an OA prevalence of 38.6% at an average 7.86 year follow up. As a result, it is estimated that osteoarthritis secondary to ACL injury leads to 25,000 to 30,000 total knee arthroplasties per year (Mather III et al. 2013). Therefore, it is obvious that ACL injury is a very serious, detrimental, and life altering issue.

2.3 ACL Injury Mechanism: Of the ways to injure the ACL, a sudden deceleration during landing or cutting is a commonly reported mechanism (Boden et al 2000, Olsen et al 2004). This is due to excessive loading of the knee in all three planes (frontal, sagittal, and transverse) causing damage to the internal structures of the knee (McLean et al 2004, Quatman et al 2010, Shimokochi and Shultz 2008). In a recent systematic review, Quatman et al (2010) found 32% of the diagnostic studies included supported a sagittal plane only theory for ACL injury. However, those studies only examined anterior tibial translation which occurs in the sagittal plane and does not bring in the influence of other planar motion. ACL strain has been found to be greatest under multi-planar loading compared to solely sagittal plane loading (Markolf et al 1995). Further, non-contact ACL injuries are reported to be more likely during landings with lesser knee flexion angles combined with either transverse plane knee motion, frontal plane
knee motion, or both (Quatman et al 2010, Shimokochi and Shultz 2008). While sagittal plane mechanics are important, frontal-plane motion is generally considered a critical piece in the ACL injury mechanism. Olsen et al (2004) found that abduction of the knee during landing was the common mechanism for ACL injury when examining handball players who had suffered an ACL injury. Imaging of the knee after ACL injury demonstrates bone bruising on the lateral femoral condyle 80% of the time, which is theorized to occur due to the medial joint opening and the lateral side compressing as the knee abducts (Quatman et al 2010). Further, Markolf et al (1995) examined ACL loading of cadaveric specimens under different planes of movement and reported that one of the most important loading conditions, in terms of increasing ACL strain, was anterior tibial shear force combined with an external knee abduction moment. Therefore, while ACL injury is most likely caused by a multi-planar movement, frontal-plane biomechanics likely play a crucial role in ACL injury mechanism. Thus, understanding how to prevent excessive frontal plane knee motion and loading is crucial.

2.4 Excessive Frontal Plane Knee Motion and Loading: While non-contact ACL injury likely involves knee loading in multiple planes, excessive frontal-plane knee motion is often cited as a key feature of the injury event (Olsen et al 2004, Quatman et al 2010). Hewett et al (2005) examined 205 female athletes who were screened using three dimensional motion analysis before their competitive seasons. Researchers then followed the subjects through their seasons to determine if an ACL injury occurred, as documented by their athletic trainer and confirmed by MRI or arthroscopic surgery. They found both peak knee abduction angle and external knee abduction moment were significant predictors of future ACL injury events. Hewett et al (2005) also found that knee abduction at initial contact and peak knee abduction
angles during landing were significantly different in females that went on to injure their ACL compared to those that did not. Due to these findings, extensive research has been geared towards examining factors related to frontal plane knee motion and loading (Willey and Davis 2004, Hollman et al 2013, Cashman 2012).

The hip articulation consists of the acetabulum of the pelvis and the proximal femur, meaning that motion at the hip influences femoral position which can impact knee biomechanics. During landing, the hip will commonly flex, adduct, and internally rotate (Jacobs et al 2007). Excessive hip adduction, while the foot is planted on the ground, can lead to medial excursion of the knee joint center that positions the knee in an abducted position (Powers 2010). When examining the influence of hip mechanics on knee injury, Powers (2010) found that individuals with hip abductor weakness may shift their weight over the stance limb in order to compensate for the weakness. They proposed that this movement pattern, described as a “compensated Trendelenburg sign”, places increased stress on the medial portions of the knee and could potentially lead to an increased knee abduction angle. During landing on a single leg, those compensations will contribute to move the ground reaction force vector lateral, thereby exaggerating the amount of frontal plane knee motion (Powers 2010). This exaggerated knee abduction angle can increase frontal plane knee loading which could potentially place greater stress on the ACL (McLean et al 2004, Markolf et al 1995).

2.5 Role of the Hip Musculature

2.5.1 Hip Muscle Function: The relationship between hip muscle function and knee abduction is thought to be particularly important. Greater knee abduction angles have been shown to be associated with greater hip adduction and internal rotation angles (Padua et al
2005). As the gluteus maximus and medius are two of the larger muscles that can work eccentrically to control these motions during landing (Neumann 2010), these muscles are important to consider as a means of controlling excessive frontal plane knee motion.

When examining factors associated with increased risk of ACL injury, decreased hip abductor peak strength has been correlated with increased peak knee abduction angles and moments (Jacobs and Mattacola 2005, Powers 2010). Increased strength of the gluteus maximus has also been associated with lesser knee abduction (Hollman et al 2009, Hollman et al 2013). The gluteus maximus is the largest muscle of the hip and due to its orientation and cross-sectional area has the greatest force producing capacity for external rotation compared to the other primary external rotators of the hip (Neumann 2010). Research has also found the gluteus medius has the greatest moment arm for producing abduction and accounts for 60% of the total hip abductor cross-sectional muscle area, compared to 40% in the gluteus minimus and tensor fascia latae combined (Neumann 2010). During landing, the hip tends to be in a relatively adducted and internally rotated position (Powers 2010), which means the hip abductors and external rotators are elongated, putting them in a position which can improve their force producing capacity (Neumann 2010). A weakness of the hip abductors has been theorized to lead to greater hip adduction angles during landing (Powers 2010). Similarly, it is plausible that weakness of the gluteus maximus and external rotators could result in a decreased ability to control femoral internal rotation and subsequent knee abduction. As such, the hip abductors, extensors, and external rotators likely play a vital role in controlling frontal plane knee motion.
2.5.2 Hip Abductor and Extensor Peak Torque: The majority of research pertaining to hip muscle function and its relationship to frontal plane knee motion has focused on peak strength (Jacobs and Mattacola 2005, Hollman et al 2009, Hollman et al 2013, Homan et al 2013, Stearns et al 2013, Stearns and Powers 2014, Willy and Davis 2011). Stearns and Powers (2014) sought to examine the effect of strengthening the hip extensors and hip abductors on frontal-plane knee biomechanics measured during double leg landings. Following a four week training program, the peak isometric strength of the hip extensors and abductors increased, while the average internal knee adductor moment during landing decreased. There was also a trend towards decreased peak knee abduction angle ($p = 0.07$), with a mean difference of 1.2 degrees (6.8 to 5.6 degrees) between pre- and post-testing. While the results demonstrate a trend toward hip strength influencing frontal plane knee motion, the magnitude of the changes observed was relatively small. However, this could have been due to the task demand, as the researchers had the subjects perform double leg landings as opposed to single leg landings.

The use of a more difficult single leg landing task may have resulted in greater knee abduction angles pre-training such that the change in knee abduction angle following the strengthening intervention may have been larger and achieved statistical significance. Jacobs and Mattacola (2005) reported a trend ($p = 0.07$, Effect size = 0.62) for women to have greater peak knee abduction angles compared to men during single leg landing, but found no significant difference in eccentric peak torque. This lack of sex differences in peak strength despite the trend for greater peak knee abduction in females might be explained by the testing position used for the gluteus medius. Researchers had the subjects perform the eccentric hip adduction in a standing position which could lead to more work for the contralateral side stabilizing the
body as opposed to the side being tested. However, these results may also be explained by the limitations of using peak strength as the measure by which to explain motion (knee abduction) that occurs very quickly after landing.

While the gluteus maximus is primarily a hip extensor and external rotator, it can assist the gluteus medius with hip abduction (Neumann 2010). Hollman et al (2009) found increased gluteus maximus activation is associated with decreased knee abduction. However, this study also reported that greater hip abduction strength was associated with greater hip adduction during a single leg step-down; which is contrary to the basic function of the hip abductors (Neumann 2010). Hollman et al (2009) postulated that a reason for the unexpected findings is that the gluteus medius, the primary hip abductor, also has a secondary action as a hip internal rotator as the hip flexes. The real flaw, however, may be found in examining the methods used to calculate the MVIC for hip abduction. The hip was placed in 30 degrees of abduction, along with slight extension. However, the step-down action analyzed involved the hip starting in neutral and moving into hip adduction. The researchers may have found different results if they had performed the strength testing in a more applicable position. The gluteus medius produces much more torque at an elongated position compared to a shortened position of 30 degrees (Neumann 2010). Another potential reason for the Hollman et al (2009) results relates to the task they chose, as a single-leg step down is not particularly demanding. Therefore, those with strength deficits could have easily compensated during the action and confounded the results. The use of a more demanding task may have potentially resulted in different findings.

A similar study was performed by Hollman et al. (2013) who examined the relationship between peak gluteus maximus strength and frontal-plane knee motion during jump landings.
The researchers found lesser hip extensor strength was correlated to greater knee abduction angles. They note both peak strength and muscle recruitment are important due to the dynamic nature of a jump-landing task and that improving muscle recruitment in conjunction with increasing peak strength is critical (Hollman et al. 2013). Hollman et al. (2013) also found increased hip internal rotation to be most correlated to increased knee abduction. This is believed to be secondary to hip muscle weakness since the gluteus maximus is also a primary hip external rotator. However, a limitation of this study is that it did not take into account the role of hip abductor strength, which has been shown to be related to increased knee abduction (Stearns and Powers 2014, Jacobs and Mattacola 2005).

As demonstrated in this review, the gluteus maximus and gluteus medius play a crucial role in influencing the magnitude of frontal plane knee motion and loading by controlling hip adduction and internal rotation. However, most previous research has only examined the influence of peak strength or torque of the hip muscles on frontal plane knee biomechanics and with inconsistent results. One potential reason for the equivocal findings could be the way in which strength was defined and assessed may not be representative of what is required of these muscles during dynamic landing tasks.

### 2.6 Rate of Torque Development:

In sport, the use of rapid torque to produce explosive movements is a part of athletic performance (de Ruiter et al. 2006). However, when dealing with injury prevention, rapid torque production might be key in activating muscles in order to control motions that can stress joints. The prevention of excessive frontal plane knee motion requires the muscles of the hip to rapidly contract over a short period of time in order to control hip adduction and internal rotation. However, during landing, maximum hip
adduction can be reached within 130-150 ms (LePhart et al 2002). Further, the ACL is reportedly injured within the first 60 ms of landing (Kernozek and Regan 2008, Koga et al. 2010). Koga et al (2010) estimated that the maximum force placed on the knee during single-leg landing and cutting tasks was just 40 ms after ground contact. Video analysis also found an abduction displacement of the knee from neutral at initial contact to 12 degrees of knee abduction after only 40 ms. Thus, rather than just producing a large magnitude of joint torque, it is apparent that it may be critical that the muscles about the hip be able to rapidly produce sufficient torque in a short period of time in order to control joint motions.

Peak torque development during isometric contractions can take between 250 and 400 ms (Aagaard 2003, Aagaard and Andersen 1998) with eccentric torque having a longer rate of torque production compared to concentric torque development (Aagaard and Andersen 1998). The rate of torque development (RTD) is found by looking at a torque-time curve and calculating the slope from the change in torque over the change in time (Aagaard et al 2002). Fast activities such as sprinting or jumping require a shorter duration to produce force, usually 50-200 ms (Aagaard et al 2002) compared to producing maximal force (250-400 ms).

Holtermann et al (2007) examined the RTD of subjects after isometric resistance training of the plantarflexors from onset of contraction to the time of peak torque. The researchers found as the resistance training was completed both the subjects MVIC and RTD increased. The researchers also incorporated investigation into different sets of instructions for the subjects; performing the task “as hard as possible” and performing the task “as hard and fast as possible” (Holtermann 2007). There was an increase in RTD when performing the trials under the “as hard and fast as possible” instruction, but this did not have a positive effect on maximal force.
Maximum hip adduction is reached within 150 ms (Leiphart et al 2002), but maximal torque is not achieved until at least 250 ms (Aagaard 2003, Aagaard and Andersen 1998). Also, resistance training has been shown to increase neural drive almost equally over the 0-200 ms range (Aagaard et al 2002). As such, the capacity to produce torque rapidly over a critical time period which also captures the full frontal-plane hip motion (i.e. 200 ms) may be more closely related with controlling rapid joint motion during landing than peak torque.

As seen above, peak torque has been a commonly studied factor in many studies examining the relationship between knee abduction and hip strength with equivocal results. However, it is clear that the time period following landing during which the ACL is at risk is shorter than the time it takes for peak torque to be reached. Therefore, it is plausible that examining the relationship between the RTD of the hip abductors, extensors, and external rotators and landing biomechanics related to ACL injury risk might be more useful than examining peak torque.

2.7 Landing Task: Landing on a single-leg and cutting are commonly reported mechanisms for a non-contact ACL injury. It has also been stated that previous research has sought to examine the role of the hip in controlling motion at the knee to prevent such injury. However, one potential flaw with some of this research is the choice of task that subjects completed. In their investigations related to hip strength, Hollman et al (2013), Homan et al (2013), and Stearns and Powers (2014) used double leg landings. Unfortunately, a double leg task reduces the demand on each leg to decelerate and stabilize the body compared to a single leg landing. Performing a double leg landing task is also not representative of the common mechanism for non-contact ACL injury. The researchers above may have found greater frontal
plane involvement had the subjects been required to land on one leg. In support of this notion, Homan et al (2013) used a double leg landing and found no difference in frontal plane hip and knee kinematics between groups characterized as having high or low peak abduction torque. However, they did find greater EMG activation amplitude of the hip abductors in the high versus low peak torque group. This suggests that weaker individuals activated more in order to have a comparable landing profile to the strong group, while the strong group did not require as much activation to land safely. Had the task been more difficult, such as on a single leg, then there may have been significant differences in the kinematic data between the groups.

Jacobs and Mattacola (2005) had subjects perform a single leg hop and land task and found greater mean peak knee abduction angles compared to other research done using double leg landings: 14.3 degrees vs. 3.0 degrees (Hollman et al 2013). Clairborne et al (2006) also used a single leg task when examining the relationship between peak hip and knee strength and knee abduction angle. They found that peak hip abduction torque had a weak to moderate relationship with knee abduction displacement, but was not a significant predictor of peak knee abduction angle during a single leg squat. Researchers may have found more powerful results had the task been more strenuous than five to seven single leg squats. Such a task seems very unlikely to cause frontal plane demands similar to that seen in single leg landing tasks that more closely mimic reported non-contact ACL injury mechanisms.

As seen above, the use of different tasks may contribute to the inconsistency of the results in studies examining hip strength and landing mechanics. This is potentially due to a lack of stress on the joint, allowing subjects to alter their kinematics without concern for potential consequences. As such, when looking to examine the kinematics related to ACL injury
it is important consider a task which will be appropriately strenuous.

2.8 Conclusion: ACL injury has serious short term consequences, months of recovery and rehabilitation before return to play, and significantly increases the chances of developing early-onset OA. Even with so much research done on the factors contributing to such an injury, there has been no appreciable change in the injury rate in the United States. As the ACL is placed under greater strain with increased frontal plane knee loading, research has been geared towards identifying ways to decrease knee abduction during landing with proximal control of femoral motion via the hip abductors, extensors, and external rotators an active area of investigation. However, most of this previous research has examined how peak hip abductor, extensor, and/or external rotator strength influences frontal plane hip and knee biomechanics. With ACL injury occurring within the first 60 ms of landing, the assessment of RTD during a time critical period (200 ms) is a novel concept for evaluating whether the ability to rapidly produce torque may be more closely linked to control of frontal plane motion during dynamic tasks than peak torque. Therefore, the purpose of this study was to examine the influence of rapid torque production of the hip abductors, extensors, and external rotators on frontal plane hip and knee biomechanics during a single-leg jump-cut task.
Chapter 3: Methods

3.1 Subjects: Forty, healthy, recreationally active female volunteers between 18-30 years old (age = 21.0 ± 1.7 years; height = 67.4 cm ± 7.81 cm; mass 65.89 kg ± 8.54 kg) were recruited from the local college population and surrounding Corvallis area as participants in this investigation. Recreationally active was defined as participating in at least 150 minutes of moderate to vigorous physical activity per week (Garber et al 2011). All participants also reported: 1) no history of lower extremity surgery; 2) no current injury or illness which limits their physical activity level; 3) no leg or low back injury in the last 6 months which limited their physical participation; 4) no previous ACL injury, and 5) that they had participated in an activity involving cutting or jumping within the last 6 months. Subjects were also asked not to partake in any strenuous exercise 24 hours prior to the testing session.

3.2 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. All participants were outfitted in spandex shorts and tank top and wore their own athletic shoes during testing. The height and mass of each subject was recorded prior to data collection for biomechanical model generation and standardization of the dependent variables. Leg dominance was determined by which leg subjects used at least twice for the following tasks: 1) to kick a ball for distance, 2) to step onto a 25cm box, and 3) to recover from small perturbation from behind. Participants then completed a 5 minute warm-up at moderate intensity on a stationary bike.

3.2.1 Rate of torque development: Rate of torque development was measured using a Biodex System 3 dynamometer (Biodex Medical Systems Inc., Shirley, New York, USA) interfaced with
The MotionMonitor motion analysis software (Innovative Sports Training, Chicago, IL, USA). Participants performed three maximum isometric voluntary contractions (MVIC) of the hip abductors, followed by the hip extensors, and then the hip external rotators. For each MVIC trial subjects were instructed to contract “as hard and fast as possible” for three seconds in response to a light stimulus. Participants completed three MVIC trials for each muscle group with one minute of rest between trials. Torque-time curves from each MVIC trial were immediately evaluated for an initial countermovement and a plateau on the curve. Trials exhibiting a countermovement and/or no plateau were not used and the trial repeated until a maximum of five trials were completed.

**Hip abduction (Figure 1):** The subject was positioned in a side-lying position on the non-dominant side with the hip center of rotation aligned with the dynamometer axis of rotation. The hip was stabilized with a strap over the iliac crest and the resistance pad of the movement arm was placed just superior to the lateral epicondyle. The subject was strapped with the test leg in 15 degrees of extension, with zero degrees of hip abduction and external rotation. The subject was instructed to push their leg straight up into the resistance pad.

**Hip extension (Figure 2):** The subject was placed prone with the legs off the end of the Biodex. The movement arm was placed just superior to the popliteal fossa. The greater trochanter was aligned with the dynamometer axis of rotation. Using a goniometer, the test leg was placed in 30 degrees of hip flexion and 90 degrees of knee flexion. The subject wrapped their arms around the chair and the low back was strapped down. The subject was instructed to kick their thigh straight backwards while maintaining 90 degrees of knee flexion.

**Hip external rotation (Figure 3):** The subject was seated with the legs off the edge of the Biodex.
The movement arm was placed just superior to the medial malleolus. The middle of the patellar tendon was aligned with the dynamometer axis of rotation. Using a goniometer, the test leg was placed in 90 degrees of hip flexion and 90 degrees of knee flexion. The subjects test leg and hips was strapped down and arms crossed across the chest. The subject was instructed to turn their lower leg towards their opposite lower leg.

### 3.2.2 Lower Extremity Biomechanics Assessment:

Following RTD testing, lower-extremity biomechanics were assessed during a single-leg jump-cut task. Participants were outfitted with a retro-reflective marker set, (27 static, 23 dynamic) 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 the sacrum. Also, markers were placed bilaterally on the shoes over the approximate locations of the calcaneus, and the 1st and 5th metatarsal heads. Nine motion capture cameras (Vicon, Inc., Centennial, CO, USA) were used to record participant kinematics during single-leg jump-cuts following a static subject calibration trial. Single-leg jump-cuts were performed from a distance equal to 50% of the subject’s height away from the edge of a force plates (Type 4060-08, Bertec Corporation, Columbus, OH, USA). Participants performed the jump-cut task as described by Frank et al (2013). Participants stood a distance of 50% of their body height from the nearest edge of the force plate and a hurdle 17cm high was placed at 25% of their body height in front of the force plate. The participant performed a double leg jump over the hurdle, landed using only the dominant leg with the foot positioned entirely on the force plate, and then cut at a 60 degree angle in the direction opposite the dominant leg as fast as possible (i.e. landing on the right leg required a cut to the left). Participants performed at least three practice trials after
receiving the instructions before completing five testing trials.

3.3 Data Sampling, Processing, and Reduction

3.3.1 RTD: The raw voltage signal from the Biodex System 3 dynamometer was sampled at 1560 Hz using The MotionMonitor software during the isometric contraction trials. The recorded voltage signal for each trial was digitally low-pass filtered at 10 Hz using a fourth-order Butterworth filter and converted to torque (Nm) via a calibration equation function using custom written computer software (LabVIEW, National Instruments Corp., Austin, Texas). This same custom software was also used to calculate RTD during each MVIC trial during the 200 ms after the onset of muscle contraction (i.e., time point when torque exceeded 2.5% of peak torque). RTD was calculated as the slope of the line that best fits the torque-time curve over the period of interest (0-200 ms). The maximum RTD value across the three trials for each motion was normalized to the product of subject height and weight and used for statistical analysis.

3.3.2 Lower Extremity Biomechanics: The kinematic and force plate data during the single-leg jump-cut task were sampled at 120 and 1560 Hz, respectively, using the Vicon Motion Analysis software (Vicon, Inc., Centennial, CO, USA). Raw three-dimensional kinematic coordinates and kinetic data were imported into The Motion Monitor motion analysis software for biomechanical model generation (Innovative Sports Training, Chicago, IL, USA). Ankle and knee joint centers were defined as the midpoint of the medial and lateral malleolus and the medial and lateral femoral epicondyle markers, respectively. The hip joint center was predicted using external landmarks as described by (Bell et al 1989). The local coordinate systems of the shank, thigh, and sacrum were defined with the positive x-axis directed anteriorly, positive y-axis
directed to the left, and the positive z-axis directed superiorly. Kinematic and force plate data were low-pass filtered at 10 Hz using a 4th order zero-phase lag Butterworth digital filter, and kinematic data was time-synchronized to kinetic data, and re-sampled at 1560 Hz. Joint angular positions were calculated based on a right hand convention using Euler angles in a Y (flexion/extension), X’ (adduction/abduction), Z” (internal/external rotation) rotation sequence with motion defined about the knee as the shank relative to the thigh and about the hip as the thigh relative to the pelvis. Frontal plane inter-segmental knee moment of force was calculated using an inverse dynamics solution within The MotionMonitor using the methods described by Gagnon and Gagnon (1992). Custom computer software (LabVIEW Inc.) was used to identify dominant limb peak knee abduction angle, knee abduction angle at initial contact, knee abduction displacement, peak hip adduction angle, hip adduction angle at initial contact, hip adduction displacement, and peak internal knee adduction moment for each trial between initial contact and the time of peak knee flexion. Peak internal knee adduction moment was normalized to the product of subject height and weight. All dependent variables were averaged across the five trials prior to statistical analysis.

3.4 Statistical Analysis: Following data reduction, subjects were divided into tertiles for each of the three RTD measures (abduction, extension, and external rotation) independently. All data were then checked for consistency and tested for violations of the assumption of normality. Separate one-tailed, independent samples t-tests or Mann-Whitney U tests were then used as appropriate to identify significant differences in the dependent variables between the highest (High) and lowest (Low) tertiles for each RTD measure as summarized in Table 1. All statistical procedures were performed using IBM SPSS Statistics 23.0 (IBM, Armonk, New York, USA).
### Table 1: Research Questions, Independent Variables, Dependent Variables and Statistical Analysis Plan.

<table>
<thead>
<tr>
<th>Research Question</th>
<th>Independent Variable</th>
<th>Dependent Variable</th>
<th>Test</th>
</tr>
</thead>
</table>
Figure 1: Subject position for RTD of the hip abductors

Figure 2: Subject position for RTD of the hip extensors
Figure 3: Subject position for RTD of the hip external rotators
Chapter 4: Results

The allocation of participants into tertiles was successful in creating groups with significantly different rates of torque development for hip abduction, extension, and external rotation RTD (Table 2). No significant differences in frontal plane hip or knee biomechanics were identified between High and Low hip abduction RTD groups (Table 3). For hip extension RTD, the High group had greater peak hip abduction angle during landing (-8.0° vs -3.3°), and also went through greater hip adduction (6.5° vs 3.8°) and knee abduction (-4.4° vs -2.3°) displacements than the Low hip extension RTD group. However, no differences in frontal plane hip or knee angles at initial contact or peak knee abduction angle or peak internal knee adduction moment were identified (Table 4). Finally, for hip external rotation RTD, the High group had greater hip adduction (8.0° vs 5.2°) and knee abduction (-4.5° vs -2.4°) displacements than the Low RTD group. No other significant differences in frontal plane hip or knee biomechanics were identified (Table 5).
Table 2: Group comparison of Hip Abduction, Extension, and External Rotation RTD

<table>
<thead>
<tr>
<th></th>
<th>Hip Abduction RTD (Nm x [BW*Ht]⁻¹)</th>
<th>Hip Extension RTD (Nm x [BW*Ht]⁻¹)</th>
<th>Hip External Rotation RTD (Nm x [BW*Ht]⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low</td>
<td>0.31 ± 0.07</td>
<td>0.31 ± 0.08</td>
<td>0.12 ± 0.02</td>
</tr>
<tr>
<td>High</td>
<td>0.62 ± 0.13</td>
<td>0.68 ± 0.11</td>
<td>0.22 ± 0.02</td>
</tr>
<tr>
<td>p-value</td>
<td>&lt; 0.001</td>
<td>&lt; 0.001</td>
<td>&lt; 0.001</td>
</tr>
</tbody>
</table>

Significant difference between groups when \( p < 0.05 \)

Table 3: Hip Abduction RTD Group Comparison of Frontal Plane Hip and Knee Biomechanics

<table>
<thead>
<tr>
<th></th>
<th>Hip Adduction at Initial Contact (°)</th>
<th>Peak Hip Adduction (°)</th>
<th>Hip Adduction Displacement (°)</th>
<th>Knee Abduction at Initial Contact (°)</th>
<th>Peak Knee Abduction (°)</th>
<th>Knee Abduction Displacement (°)</th>
<th>Peak Knee Adduction Moment (Nm x [BW*Ht]⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low</td>
<td>-12.6 ± 6.9</td>
<td>-6.1 ± 7.3</td>
<td>6.4 ± 4.0</td>
<td>-3.2 ± 3.6</td>
<td>-8.5 ± 5.6</td>
<td>-5.3 ± 3.6</td>
<td>0.010 ± 0.01</td>
</tr>
<tr>
<td>High</td>
<td>-10.2 ± 4.4</td>
<td>-4.6 ± 5.4</td>
<td>5.6 ± 3.4</td>
<td>-1.7 ± 2.1</td>
<td>-5.3 ± 4.9</td>
<td>-3.6 ± 4.2</td>
<td>0.006 ± 0.08</td>
</tr>
<tr>
<td>p-value</td>
<td>0.159</td>
<td>0.272</td>
<td>0.291</td>
<td>0.099</td>
<td>0.065</td>
<td>0.138</td>
<td>0.687</td>
</tr>
</tbody>
</table>

Abduction (-) and adduction (+) by convention
Significant difference between groups when \( p < 0.05 \)
Table 4: Hip Extension RTD Group Comparison of Frontal Plane Hip and Knee Biomechanics

<table>
<thead>
<tr>
<th></th>
<th>Hip Adduction at Initial Contact (°)</th>
<th>Peak Hip Adduction (°)</th>
<th>Hip Adduction Displacement (°)</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Low</strong></td>
<td>-9.8 ± 5.9</td>
<td>-3.3 ± 5.7</td>
<td>6.5 ± 3.0</td>
<td>0.150</td>
</tr>
<tr>
<td><strong>High</strong></td>
<td>-11.8 ± 3.8</td>
<td>-8.0 ± 4.5</td>
<td>3.8 ± 3.0</td>
<td>0.015</td>
</tr>
<tr>
<td><strong>p-value</strong></td>
<td>0.150</td>
<td>0.015</td>
<td>0.019</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>Knee Abduction at Initial Contact (°)</th>
<th>Peak Knee Abduction (°)</th>
<th>Knee Abduction Displacement (°)</th>
<th>Peak Knee Adduction Moment (Nm x [BW*Ht]⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Low</strong></td>
<td>-3.0 ± 3.5</td>
<td>-7.4 ± 4.8</td>
<td>-4.4 ± 3.2</td>
<td>0.007 ± 0.008</td>
</tr>
<tr>
<td><strong>High</strong></td>
<td>-2.3 ± 2.6</td>
<td>-4.8 ± 3.9</td>
<td>-2.5 ± 2.3</td>
<td>0.007 ± 0.008</td>
</tr>
<tr>
<td><strong>p-value</strong></td>
<td>0.281</td>
<td>0.071</td>
<td>0.046</td>
<td>0.920</td>
</tr>
</tbody>
</table>

Abduction (-) and adduction (+) by convention
Significant difference between groups when p < 0.05

Table 5: Hip External Rotation RTD Group Comparison of Frontal Plane Hip and Knee Biomechanics

<table>
<thead>
<tr>
<th></th>
<th>Hip Adduction at Initial Contact (°)</th>
<th>Peak Hip Adduction (°)</th>
<th>Hip Adduction Displacement (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Low</strong></td>
<td>-13.0 ± 5.9</td>
<td>-4.9 ± 5.8</td>
<td>8.0 ± 3.3</td>
</tr>
<tr>
<td><strong>High</strong></td>
<td>-10.3 ± 4.6</td>
<td>-5.1 ± 6.0</td>
<td>5.2 ± 3.3</td>
</tr>
<tr>
<td><strong>p-value</strong></td>
<td>0.105</td>
<td>0.472</td>
<td>0.019</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>Knee Abduction at Initial Contact (°)</th>
<th>Peak Knee Abduction (°)</th>
<th>Knee Abduction Displacement (°)</th>
<th>Peak Knee Adduction Moment (Nm x [BW*Ht]⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Low</strong></td>
<td>-2.1 ± 3.6</td>
<td>-6.7 ± 5.2</td>
<td>-4.5 ± 3.2</td>
<td>0.007 ± 0.012</td>
</tr>
<tr>
<td><strong>High</strong></td>
<td>-2.1 ± 3.7</td>
<td>-4.5 ± 5.7</td>
<td>-2.4 ± 2.9</td>
<td>0.009 ± 0.016</td>
</tr>
<tr>
<td><strong>p-value</strong></td>
<td>0.487</td>
<td>0.162</td>
<td>0.044</td>
<td>0.362</td>
</tr>
</tbody>
</table>

Abduction (-) and adduction (+) by convention
Significant difference between groups when p < 0.05
Chapter 5: Discussion

The primary finding of this investigation is that while hip abductor RTD was not related to frontal plane hip or knee biomechanics, females with greater hip extension RTD or greater external rotation RTD exhibited lesser hip adduction and knee abduction displacements when performing a single-leg jump-cut.

Contrary to our hypothesis, the capacity to rapidly generate hip abduction torque did not influence frontal plane hip or knee kinematics (initial contact, peak, or displacement), or peak internal knee adduction moment during a single-leg jump-cut task. While unexpected, one potential reason for the lack of association is that the single-leg jump-cuts resulted in participants being in a relatively abducted hip position as they landed and changed direction. On average, subjects were positioned in an abducted hip position at initial contact and remained abducted during the entire loading phase. As a result, the gluteus medius, which acts as the primary hip abductor, was in a suboptimal position to eccentrically control hip adduction (Neumann 2010). While the gluteus medius has the greatest force producing capacity of all hip abductors, its ability to produce force decreases in a nearly linear fashion as the hip moves from 10° of adduction and into an abducted position (Dostal et al 1986, Neumann 2010, Neumann et al 1988, Olson et al 1972).

It is also important to note that the use of an abducted hip position is not isolated to the jump-cut task used in this study, but has been observed in a variety of single leg landing and cutting movements. Houck et al (2006) reported an average of 10.6° of hip abduction during an anticipated side-step cutting task. Sigward and Powers (2007) compared females with excessive knee valgus at landing with a control group and reported an average of 12.8° and 7.7
of hip abduction, respectively, during a 45° side-step cut. These findings are consistent with our research, which indicates that subjects generally landed with between 10-13° degrees of hip abduction, and supports the idea that the completion of a single-leg cutting task inherently requires a relatively abducted, instead of adducted, hip position.

Along with hip abduction, greater hip flexion also serves to decrease the force producing capacity of the gluteus medius. As the hip flexes from zero to 40°, the moment arm vector component of the gluteus medius for hip abduction decreases, resulting in a reduction in the effectiveness for hip abduction torque production (Dostal et al 1986). In addition to performing single-leg jump-cuts in an abducted position, secondary analysis of our data indicated that, on average, participants were in about 30° of hip flexion at initial contact and throughout the loading phase of the single-leg jump-cut. The combined use of an abducted and flexed hip likely served to significantly reduce the capacity of the gluteus medius to eccentrically control hip adduction and may explain why participants with greater hip abduction RTD did not exhibit differences in frontal plane hip and knee biomechanics compared to those with lesser explosive hip abduction strength.

In contrast, we identified that greater hip extension RTD or greater external rotation RTD were related to lesser hip adduction and knee abduction displacements. Further, it is not surprising that we observed similar results for hip extension and external rotation RTD due to the importance of the gluteus maximus for both of these actions. The gluteus maximus is the largest and most powerful hip extensor (Delp et al 1999, Neumann 2010), but it is also the largest hip external rotator (Neumann 2010). Given that external rotation RTD was measured with our subjects in 90 degrees of hip flexion, which diminished the effectiveness of the short
lateral hip external rotators by moving the moment arm of these muscles moving towards an internal rotation line of action (Delp et al 1999), it is likely that both hip extension and external rotation RTD, as measured in this investigation, are reflective of explosive strength of the gluteus maximus.

While the gluteus maximus is primarily, and accurately, viewed as a hip extensor and external rotator, there is some evidence that the upper portion of the gluteus maximus functions as a hip abductor, with this action becoming more pronounced as the hip becomes more flexed. Shen (1982) reported that in children with gluteus maximus contracture, the hips will be held in an extended position while standing. However, when patients are seated and the hips flexed, the thighs will become fixed in an abducted position suggesting that at least some portion of the gluteus maximus has an abduction line of action (Shen 1975). Lyons et al (1983) reported that the activation amplitude and timing of the upper portion of the gluteus maximus mirrored the gluteus medius during free walking and stair ascent and suggested that the upper gluteus maximus acts as a hip abductor during the loading response and during single leg support. Moreover, in those with advanced hip osteoarthritis, the upper portion of the gluteus maximus on the affected side has been shown to atrophy, while the unaffected side actually hypertrophies due to the gluteus maximus’s role as a hip abductor in preventing contralateral hip drop (Grimaldi et al 2009). Given this, and the fact that the upper portion of the gluteus maximus is located superior to the hip joint center of rotation and inserts on the IT band, it has been suggested that this portion of the gluteus maximus should be considered part of a hip abductor synergy that also includes the tensor fascia latae (Grimaldi 2011). Therefore, it is possible that during the single-leg jump-cut, the upper portion of the gluteus maximus was
able to take on an important role in controlling frontal plane hip motion in place of the gluteus medius, whose function was compromised by the abducted and flexed hip position used by participants to complete the task. This notion is supported by our data, which demonstrates that those categorized as having High gluteus maximus RTD (i.e. hip extension or external rotation) exhibited significantly lesser hip adduction displacement than those categorized with Low gluteus maximus RTD.

Along with the significant difference in hip adduction displacement between the High and Low hip extension and external rotation RTD groups, we also identified lesser knee abduction displacements in these same High RTD groups. Previous research also supports the notion that hip extensor strength and the gluteus maximus activation are related to lesser knee abduction (Hollman et al 2009, Hollman et al 2013). Hollman et al (2009) reported that increased gluteus maximus activation was associated with a decrease in peak knee abduction angles during a single-leg step down, and that greater hip extensor strength and activation were predictive of lesser peak knee abduction angle during a jump-landing task (Hollman et al 2013). While the difference in knee abduction displacement was statistically significant, the magnitude of difference between groups was only about two degrees and was not accompanied by a significant group difference in frontal plane knee loading. The minimal magnitude of differences in knee abduction displacement coupled with a lack of differences in knee joint loading between High and Low hip extension and external rotation groups suggests that the task was not demanding enough for the increased frontal plane hip and knee motion to manifest in greater frontal plane knee joint loading. However, it is possible that in more demanding cutting tasks like those encountered in sporting activities, individuals with lesser hip
extensor or external rotator (i.e. gluteus maximus) RTD might experience greater frontal plane motion and knee loading, thereby increasing the risk for ACL injury.

As with frontal plane hip kinematics, we found no relationship between hip abduction RTD and frontal plane knee kinematics. Similarly, Homan et al (2013) reported no difference in peak knee abduction angles between groups classified as having strong (15.9°) and weak (17.8°) hip abductors. A systematic review of seven studies evaluating the relationship between hip abductor strength and frontal plane knee motion, only one identified a significant relationship between greater hip abductor strength and lesser peak knee abduction angle or knee abduction displacement (Cashman 2012). The results of these studies, coupled with the results of this investigation, indicates that hip abductor strength may not play as significant a role in controlling frontal plane hip and knee motion and loading during all types of movement tasks as has been previously proposed (Jacobs and Mattacola 2005).

While results related to hip abduction RTD and frontal plane hip and knee biomechanics were unexpected, this investigation presents novel findings related to the importance of the gluteus maximus that have clinical relevance. The results regarding hip extension and external rotation RTD highlight the potential importance of explosive gluteus maximus strength in controlling frontal plane motion at the hip and knee, and suggest that including activities to increase explosive strength of this muscle may be crucial in the prevention and rehabilitation of a variety of knee and hip injuries. While the gluteus medius has long been believed to be the primary muscle to be targeted in rehabilitation and prevention programs in relation to reducing frontal plane knee loading, the current results do not support that paradigm during all types of movement tasks. Our findings indicate that in tasks such as jump-cuts that are performed with
the hip in an abducted and flexed position, the gluteus maximus potentially plays a large role in controlling frontal plane hip and knee motion. As a result, we propose that focused training of the gluteus maximus, with particular emphasis on increasing explosive strength through rapid muscle contractions, may be useful to improve eccentric hip control during single-leg landing and cutting maneuvers that present a significant risk of injury. However, future research is necessary to confirm this hypothesis.
Chapter 6: Conclusion

While we did not identify significant differences in frontal-plane biomechanics between groups of participants classified as having High and Low hip abductor RTD, participants with High hip extension or external rotation RTD demonstrated lesser hip adduction and knee abduction displacements during a single-leg jump-cut compared to those with Low RTD. We believe these results can be attributed to: 1) a reduction in the capacity of the gluteus medius to produce hip abduction torque when the hip is abducted and flexed as was the case during the jump-cut used in this investigation; and 2) the potential significance of the upper portion of the gluteus maximus in functioning as a hip abductor. The findings highlight the previously unreported importance that the gluteus maximus could have in controlling frontal-plane motion during sport-specific tasks such as cutting that present greater risk for ACL injury. Further, we propose that it might be beneficial for clinicians to specifically emphasize explosive strength training of the gluteus maximus in those who participate in these types of cutting activities. However, future research is necessary to corroborate our findings and to determine whether increasing explosive gluteus maximus strength actually results in more favorable frontal-plane hip and knee biomechanics during landing and cutting.
Bibliography


