THE EFFECTS OF HIP STRENGTH ON GLUTEAL MUSCLE ACTIVATION AMPLITUDES AND HOW THESE FACTORS PREDICT LOWER EXTREMITY KINEMATICS

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Abstract

KATIE J. HOMAN: The effects of hip strength on gluteal muscle activation amplitudes and how these factors predict lower extremity kinematics.
(Under the direction of Dr. J Troy Blackburn, PhD, ATC)

Objective: To evaluate the influence of hip strength on kinematic ACL injury risk factors and gluteal activation. Design: Cross-sectional. Setting: Research laboratory. Participants: Eighty-two healthy volunteers. Outcome Measures: Hip extension, external rotation, and abduction strength; gluteus maximus and medius electromyography (EMG); knee valgus, hip adduction, and hip internal rotation angles. Results: Peak knee valgus (p=0.016) and hip external rotation (p=0.023) angles were greater in individuals with weaker hip external rotators. Gluteus maximus EMG was greater in individuals with weaker hip extensors (p=0.031) and external rotators (p=0.043). Hip external rotation strength and gluteus maximus EMG amplitude predicted 7.9% and 12.1% of the variance in peak hip rotation (R=0.281, p=0.039.) and knee valgus angles (R=0.348, p=0.006). Conclusion: Hip external rotation strength influences hip and knee kinematics related to ACL injury, thus increasing gluteal strength may be an important addition to ACL injury prevention programs. Key Words: ACL, strength, EMG, knee valgus
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<th>Abbreviation</th>
<th>Description</th>
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<tr>
<td>ACL</td>
<td>Anterior Cruciate Ligament</td>
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<tr>
<td>EMG</td>
<td>electromyography</td>
</tr>
<tr>
<td>ICC</td>
<td>Intraclass Correlation Coefficients</td>
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<tr>
<td>NMRL</td>
<td>Neuromuscular Research Laboratory</td>
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<tr>
<td>MVIC</td>
<td>Maximum Voluntary Isometric Contraction</td>
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<tr>
<td>SEM</td>
<td>Standard Errors of Measurements</td>
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<td>Q-angle</td>
<td>Quadriceps angle,</td>
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Chapter 1

Introduction

Approximately 250,000 anterior cruciate ligament (ACL) tears are estimated to occur annually in the United States, resulting in multiple short and long term effects including financial, psychological, and physical stresses (Hewett, et al., 1998). Financially, conservative estimations calculate that Americans spend $1.7 billion annually repairing these injuries (Hewett, Myer, & Ford, 2006; Miyasaka, 1991). Psychologically, an ACL tear can cause an athlete to miss 6-12 months of participation, which can be emotionally challenging for an athlete. Athletes who have an injury for which they miss a significant amount of time have poorer academic performance (Freedman, Glasgow, Glasgow, & Bernstein, 1998). An ACL injury also results in degenerative changes in the knee. O’Neill (2001) found 11.6% of 225 subjects demonstrated osteoarthritis 6 to 11 years post-surgery, and Maletius and Messner (1999) found that 84% of 56 patients who had ACL surgery demonstrated mild to moderate osteoarthritis twenty years post-surgery. Due to the number, cost, and secondary effects of ACL injuries, it is important to understand factors that may predispose individuals to an ACL injury so that an effective injury prevention protocol can be created.

Most ACL injury are reported to be non-contact injuries which occur during jump landings, cutting maneuvers, deceleration, planting, or pivoting (Griffin, et al., 2000). An
extensive amount of research has been performed to identify factors that predispose individuals to an ACL injury in an effort to develop injury prevention guidelines. This research has shown that there are multiple intrinsic and extrinsic risk factors that may predispose athletes to ACL injuries. A great deal of research has been focused on females, as studies show that females are 2-8 times more likely to experience an ACL injury (Arendt & Dick, 1995; Bjordal, Arnly, Hannestad, & Strand, 1997). Categories of intrinsic risk factors associated with ACL injuries include anatomical, hormonal, biomechanical, and neuromuscular; a previous history of an ACL injury is also a potential intrinsic risk factor. Shoe to surface interface and contact with another player or object are both considered extrinsic risk factors for an ACL injury (Boden, Griffin, & Garrett, 2000; Griffin, et al., 2000; Hewett, et al., 2006; Ireland, 1999). However, only biomechanical and neuromuscular factors are considered to be modifiable through intervention programs, thus they have received the most attention in the research literature.

One biomechanical risk factor that has been widely researched is knee valgus. Hewett et al. (2005) found a significant difference in the knee valgus angle at initial contact (p<0.01) and maximum displacement (p<0.01) between subjects that went on to tear their ACL and an uninjured group. Specifically the injured cohort had an 8.4º greater knee valgus angle at initial ground contact and 7.6º greater knee abduction displacement. Knee valgus results from a combination of femoral adduction, femoral internal rotation, tibial external rotation, and tibial abduction (Claiborne, Armstrong, Gandhi, & Pincivero, 2006; Hewett, et al., 2006). Weak musculature that contributes to these specific motions has been suggested as a potential ACL injury risk factor (Hewett, et al., 2006). It is theorized that individuals with greater hip external rotation and abduction strength may be able to resist hip adduction and internal rotation more efficiently, thus limiting knee valgus (Claiborne, et al., 2006; Hollman, et al., 2009; Russell, Palmieri-Smith, Zinder, & Ingersoll, 2006). Conversely, those with weak hip external rotators and abductors may not be able to resist hip adduction
and internal rotation as effectively, thereby contributing to greater knee valgus motion during movement and potentially greater ACL injury risk. Accordingly, it has been noted in some studies that individuals with greater hip strength display less knee valgus (Claiborne, et al., 2006; Jacobs & Mattcola, 2005; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Willson, Ireland, & Davis, 2006).

Two muscles that control hip movements associated with knee valgus are the gluteus maximus and medius. The gluteus maximus is primarily responsible for hip extension and external rotation, while the gluteus medius is primarily responsible for hip abduction (Kendall, McCreary, Provance, Rodgers, & Romani, 2005). Theoretically a weak gluteus medius muscle would result in weak hip abduction strength resulting in greater hip adduction and associated knee valgus during functional movement. A weak glutus maximus muscle would result in weak hip extension and external rotation strength resulting in greater hip internal rotation and associated knee valgus during functional movement.

The literature regarding the influence of gluteal muscle strength on knee valgus motion is inconclusive. Some studies have reported that greater knee valgus is associated with muscle weakness, while other studies have not found a significant association (Bell, Padua, & Clark, 2008; Beutler, de la Motte, Marshall, Padua, & Boden, 2009; Claiborne, et al., 2006; Jacobs & Mattcola, 2005; Jacobs, et al., 2007; Padua, et al., 2009; Sigward, Ota, & Powers, 2008; Willson, et al., 2006). Beutler et al. (2009) found that muscle strength did not strongly predict landing error scoring system (LESS) values; a qualitative tool for assessing jump-landing technique. However, females were more likely to demonstrate poor technique due to three sets of factors, including landing with greater knee valgus and a wider landing stance. Weaker hamstrings, gluteus medius, and hip internal rotators were found to be predictors of poor landing techniques in females. Unfortunately, this study did not investigate associations between strength values and specific kinematic landing patterns. Claiborne et al. (2006) found that hip abduction and internal rotation, and knee
flexion and extension strength accounted for 74.3% of the variance in knee movement in the frontal plane during a single-leg squat. This study also reported a significant negative weak-to-moderate correlation between concentric hip abduction, knee flexion, and knee extension strength and knee valgus. However, Bell et al. (2008) found that individuals with excessive medial knee displacement, which is associated with knee valgus, while performing an overhead squat had significantly greater hip extension and hip external rotation strength compared to the control group who displayed a neutral alignment. The discrepancies in the previous studies indicate the need for further researched to determine specifically if muscle strength can predict lower extremity kinematics.

There has been a considerable amount of focus on the biomechanics of the lower extremity relating to injury prevention, but neuromuscular components are starting to be researched as well. A current conjecture is that people may have the strength to dynamically control their extremity, but lack the appropriate level of muscle activation. With this said, dynamic tasks, such as landing, do not require maximal muscular force production, thus peak strength measures may not accurately predict kinematics. This indicates that the level of muscle activation may be a more important determinant of lower extremity kinematics than muscle strength (Bell, et al., 2008).

These theories may explain 1) why some athletes sustain injury while others do not and 2) the discrepancies in the literature regarding the influence of gluteal strength on knee valgus motion (Bell, et al., 2008; Claiborne, et al., 2006; Jacobs, et al., 2007; Lawrence, Kernozek, Miller, Torry, & Reuteman, 2008). Multiple factors determine muscle force including the number and timing of motor units activated to produce the muscle contraction. While a muscle may possess a high level of strength, its contributions to kinematics may be minimal if it not activated to a sufficient level. Theoretically, if two athletes have comparable muscle strength, but activate their muscles to different amplitudes, their movement patterns could potentially differ. Based on this notion, it is likely that decreased muscle strength
requires greater activation to control the lower extremity and vice versa. This suggests that both muscle strength and activation are critical contributors to lower extremity kinematics.

The current literature investigating the relationship or effect of gluteal muscle activation on knee valgus or kinematic factors associated with knee valgus is inconclusive and limited. Hollman et al. (2009) found that greater gluteus maximus activity was moderately correlated with lesser knee valgus angle during a single-leg step-down task in 20 female subjects, indicating that those with greater gluteus maximus activity demonstrated lesser knee valgus. The only other study found that specifically investigated gluteus maximus electromyography (EMG) and kinematics contradicted the previous study. Zeller, Kibler, & Uhl (2003) found no significant difference in gluteus maximus EMG activation during a single-leg squat between females and males, even though females demonstrated greater hip external rotation.

There has been more research investigating gluteus medius activation, however a lot of this research didn’t investigate kinematics. These studies contradict each other in that some found a significant difference in gluteus medius activation, while another study did not (Hanson, Padua, Blackburn, Prentice, & Hirth, 2008; Hart, Garrison, Kerrigan, Palmieri-Smith, & Ingersoll, 2007). Hanson et al. (2008) found that females had a greater gluteus medius activation compared to their male counterparts; since females experience ACL injuries at a much higher rate, this potential difference can be significant and needs to be further researched. However, the studies that have investigated gluteus medius activation and kinematics found no difference in activation amplitudes between genders even though the kinematics significantly differed (Russell, et al., 2006). Nonetheless, the contradictions in the previous literature and the limited studies that have investigated activation amplitudes and kinematics indicate the need for further research.

There have been multiple studies that have investigated how quadriceps and hamstring strength or muscle activation amplitudes affect ACL injury risk factors, but to date
there has been little research investigating how gluteal muscle strength and activation amplitudes affect these same risk factors. It is important to study these two muscles, as they primarily control hip movements associated with knee valgus. Also, most studies have reported the influence of muscle strength or muscle activation on ACL risk factors in isolation, but few studies have evaluated these variables concomitantly. With the contradicting results regarding how strength and activation amplitudes affect knee valgus, it is possible that it is a combination of these two factors that affects lower extremity kinematics, especially knee valgus. The primary purpose of this study was to investigate the influence of gluteal muscle strength on gluteal EMG amplitudes, peak knee valgus, peak hip adduction, and peak hip internal rotation during a jump landing task. A secondary purpose was to determine if the combination of gluteal strength and EMG activity predicts knee and hip landing kinematics; specifically knee valgus, hip adduction and hip internal rotation.

**Dependent/Criterion Variables**

*Kinematic Variables*

1. Peak knee valgus angle
2. Peak hip internal rotation angle
3. Peak hip adduction angle

**Independent/Predictor Variables**

*Strength Variables*

1. Hip abductor strength measured by a hand-held dynamometer
2. Hip extensor strength measured by a hand-held dynamometer
3. Hip external rotator strength measured by a hand-held dynamometer

*EMG Variables*

1. Gluteus maximus EMG amplitude
2. Gluteus medius EMG amplitude

Research Questions

*Hip Strength and Kinematics*

1. Do hip and knee kinematics during the loading phase of a double-leg jump landing task differ between groups displaying high vs. low isometric hip abduction strength?
   a. Peak knee valgus
   b. Peak hip adduction

2. Do hip and knee kinematics during the loading phase of a double-leg jump landing task differ between groups displaying high vs. low isometric hip extension strength?
   a. Peak knee valgus
   b. Peak hip internal rotation

3. Do hip and knee kinematics during the loading phase of a double-leg jump landing task differ between groups displaying high vs. low isometric hip external rotation strength?
   a. Peak knee valgus
   b. Peak hip internal rotation

*Hip Strength and EMG Amplitudes*

1. Does gluteus medius EMG amplitude during the loading phase of a double-leg jump landing task differ between groups displaying high vs. low isometric hip abduction strength?

2. Does gluteus maximus EMG amplitude during the loading phase of a double-leg jump landing task differ between groups displaying high vs. low isometric hip extension strength?
3. Does gluteus maximus EMG amplitude during the loading phase of a double-leg jump landing task differ between groups displaying high vs. low isometric hip external rotation strength?

Regressions

1. Does the linear combination of isometric hip abduction strength and gluteus medius EMG amplitude predict peak hip adduction during the loading phase of a double-leg jump landing task?

2. Does the linear combination of isometric hip external rotation strength and gluteus maximus EMG amplitude predict peak hip internal rotation during the loading phase of a double-leg jump landing task?

3. Does the linear combination of isometric hip extension strength and gluteus maximus EMG amplitude predict peak hip internal rotation during the loading phase of a double-leg jump landing task?

4. Does the linear combination of isometric hip abduction strength and gluteus medius EMG amplitude predict peak knee valgus during the loading phase of a double-leg jump landing task?

5. Does the linear combination of isometric hip external rotation strength and gluteus maximus EMG amplitude predict peak knee valgus during the loading phase of a double-leg jump landing task?

6. Does the linear combination of isometric hip extension strength and gluteus maximus EMG amplitude predict knee valgus during the loading phase of a double-leg jump landing task?
Research Hypothesis

_Hip Strength and Kinematics_

1. There will be no difference in peak knee valgus and hip adduction angles during the loading phase of a jump landing task between those demonstrating high vs. low isometric hip abduction strength.

2. There will be no difference in peak knee valgus and hip internal rotation angles during the loading phase of a jump landing task between those demonstrating high vs. low isometric hip extension strength.

3. There will be no difference in peak knee valgus and hip internal rotation angles during the loading phase of a jump landing task between those demonstrating high vs. low isometric hip external rotation strength.

_Hip Strength and EMG Amplitudes_

1. Individuals with greater isometric hip abduction strength will display lower peak gluteus medius EMG amplitudes during the loading phase of a double-leg jump landing task.

2. Individuals with greater isometric hip extension strength will display lower peak gluteus maximus EMG amplitudes during the loading phase of a double-leg jump landing task.

3. Individuals with greater isometric hip external rotation strength will display lower peak gluteus maximus EMG amplitudes during the loading phase of a double-leg jump landing task.

_Regressions_

1. The linear combination of isometric hip abduction strength and peak gluteus medius EMG amplitude will predict a significant amount of variance in peak hip adduction.
2. The linear combination of isometric hip external rotation strength and peak gluteus maximus EMG amplitude will predict a significant amount of variance in peak hip internal rotation.

3. The linear combination of isometric hip extension strength and peak gluteus maximus EMG amplitude will predict a significant amount of variance in peak hip internal rotation.

4. The linear combination of isometric hip abduction strength and peak gluteus medius EMG amplitude will predict a significant amount of variance in peak knee valgus.

5. The linear combination of isometric hip external rotation strength and peak gluteus maximus EMG amplitude will predict a significant amount of variance in peak knee valgus.

6. The linear combination of isometric hip extension strength and peak gluteus maximus EMG amplitude will predict a significant amount of variance in peak knee valgus.

**Operational Definitions**

1. Jump landing task: Subjects performed a double-leg jump landing from a 30 centimeter box placed at a distance equal to half of their height from a force plate. Subjects were required to land with their dominant foot on the force plate, their non-dominant foot completely off the force plate, and to immediately jump for max height.

2. Active subjects: Individuals who were physically actively for at least 30 minutes, 3 times per week.

3. Healthy subjects: Individuals who did not have a history of an ACL injury, back or lower extremity surgery, chronic or neurological disorders, or a lower extremity or back injury that prevented them from carrying out activities of everyday life in the past 6 months.
4. Knee valgus: the angle formed between the tibial and femoral shafts in the frontal plane
5. Muscle activation amplitude: Intensity of muscle activity as measured via EMG.
6. Loading phase: The time interval from initial ground contact to maximum knee flexion (Blackburn & Padua, 2009; Shultz, Nguyen, Leonard, & Schmitz, 2009).
7. Initial ground contact: The instant during landing at which the vertical ground reaction force measured from the force plate exceeded 10 N.
8. Dominant leg: The leg that an individual would use to kick a soccer ball for a maximum distance.

Assumptions

1. Hip extension, abduction, and external rotation strength were measured accurately.
2. Subjects carried out the task to the best of their ability.
3. Subjects were truthful about the lack of surgery, injury, or current pain.

Limitations

1. Testing took place in a laboratory setting where the subjects may not have performed the task naturally due the testing parameters including multiple wires attached to them and being observed during the task.
2. The population chosen, healthy and active individuals who are between the ages of 18 and 30 and are affiliated with the University of North Carolina, Chapel Hill, do not represent all populations, particularly those in whom ACL injury risk is greatest.
3. Only the gluteus maximus and medius muscles were tested; muscles that were not tested for EMG activation and strength could have affected kinematics.
4. All analyses were performed on the subject’s dominant leg only.
Chapter 2

Literature Review

Introduction

Conservative estimates indicate that 250,000 ACL injuries occur annually and that Americans spend $1.7 billion each year repairing these injuries (T.E. Hewett, et al., 2006; Miyasaka, 1991). Research has also shown that there has not been a significant amount of change in the occurrence of ACL injuries despite the increase knowledge about these injuries (Agel, Arendt, & Bershadsky, 2005; E. A. Arendt, Agel, & Dick, 1999; Mihata, Beutler, & Boden, 2006). Over the course of 13 years (1990-2002) there was not a significant change in the occurrence of ACL injuries in National Collegiate Athlete Association male and female soccer players (Agel, et al., 2005). The only significant difference demonstrated was the rate at which noncontact injuries in male soccer players occurred decreased. These findings suggest that while a considerable amount of research has been conducted in efforts to reduce ACL injury risk, only limited success has been achieved, thus emphasizing a continued need for research in this area.

The gross amount of ACL injuries is very alarming due to the multiple short and long term effects that transpire with this injury. Some of these short and long term effects include missing 6-12 months of participation due to recovering from surgery, financial distress, emotional distress from being out of activity for an extended period of time, and the potential for osteoarthritis. O’Neill (2001) found 11.6% of 225 subjects demonstrated osteoarthritis 6
to 11 years post-surgery. Maletius and Messner (1999) found that 84% of 56 patients who had ACL surgery had mild to moderate osteoarthritis 20 years post-surgery.

There has been a significant amount of research conducted in the past 20 years in hopes of understanding why ACL injuries occur at such an alarming rate, especially in females. Studies have been conducted to determine the epidemiology, etiology, risk factors, and potential prevention programs for this injury. Significant gains have been made in all of these topic areas, but gaps in the literature still exist. The objective of this literature review is to justify investigating the influence of gluteal muscle strength and activity on lower extremity kinematic factors. This literature review will focus on ACL injury etiology and epidemiology; biomechanical and neurological injury risk factors with a focus on knee valgus, muscle strength, and activation amplitude respectively; and the methods of a jump-landing task and EMG.

**ACL Injury Epidemiology and Etiology**

*Epidemiology*

It is estimated that females are 2-8 times more likely to tear their ACLs compared to their male counterparts, particularly in soccer, basketball, and volleyball (E. Arendt & Dick, 1995; Bjordal, et al., 1997). Agel et al. (2005) demonstrated that from 1989-2002, 682 ACL injuries occurred in basketball and 586 ACL injuries occurred in soccer out of the 15.6% NCAA schools with these two sports. Females had a 0.27 incidence rate of injuring their ACLs in basketball and 0.31 incidence rate in soccer per 1,000 exposures. Meanwhile, males had a 0.08 incidence rate in basketball and a 0.11 incidence rate in soccer per 1,000 exposures. This study represented a total of 1,008 ACL injuries in women’s basketball and soccer in over 3,392,218 exposures compared to 387 ACL injuries in 4,041,956 exposures in men’s basketball and soccer, demonstrating a higher incidence rate in females (Agel, et al., 2005).
Mihata et al. (2006) analyzed the rates of ACL tears from 1989-2004 in collegiate soccer, lacrosse, and basketball. The data was obtained from the NCAA Injury Surveillance System (ISS) in five-year increments. The injury rates for males and females in basketball and soccer were for the most part identical for each respective gender over the timeframe. For example, in women’s basketball the injury rates from 1989-1994 and 1999-2004 were 0.29 and 0.28, respectively, compared to 0.07 and 0.08 in men’s basketball during the same time periods (Mihata, et al., 2006). This pattern was similar across men’s and women’s soccer, thus supporting the data from Agel et al. (2005) and indicating that injury rates have not significantly changed over the years in basketball and soccer. The tertile values for lacrosse were not given, but over the course of 15 years, the male ACL injury rate was 0.17 and for females it was 0.18. This data indicates that females tear their ACLs at a much higher rate than males, men’s lacrosse is a much more high risk sport compared to men’s basketball and soccer, and ACL rates have not significantly changed over the course of the past fifteen years (Mihata, et al., 2006).

**Cadaver Studies**

A cadaver study performed by Berns, Hull, and Paterson (1992) suggested that anterior tibial shear force (ATSF) was the primary factor in an ACL tear. However, a knee valgus moment combined with an anterior shear force at the proximal end of the tibia created a larger strain, compared to isolated anterior shear force. Markolf et al. (1995) reported similar results, confirming that ATSF alone and knee valgus, varus, and internal rotation combined with ATSF could create enough force to damage the ACL (Markolf, et al., 1995). One study looked specifically at the effects of knee valgus on the knee and found that the load in the knee can be increased 6 times just by increasing knee valgus 5 degrees from neutral alignment (Bendijaballah, Shirazi-Adl, & Zukor, 1997). These studies signify the importance of understanding how knee valgus influences ACL injury, including what
factors predispose individuals to this kinematic motion, and what can be done to minimize this motion.

Etiology

Studies involving human subjects indicate that activities requiring cutting, sudden deceleration, pivoting, and awkward landings have been linked to ACL tears (Griffin, et al., 2000; Griffin, et al., 2006). Soccer, basketball, and lacrosse are examples of sports that meet this description. Arendt et al. (1999) demonstrated the following mechanisms of injuries in 49 ACL tears: landing from a jump (6), planting/pivoting (28), deceleration (6), going up for a jump (2), hyperextension (6), and one athlete was unsure of the mechanism. These findings demonstrate that most ACL injuries in sport occur due to a noncontact mechanism.

Risk Factors

Risk factors that can predispose athletes to ACL injuries can be classified as extrinsic or intrinsic risk factors. Extrinsic factors are those that are external or outside of the body, such as shoe to surface interface and contact with another player or object; these risk factors are usually hard to modify or to control. Intrinsic risk factors are those that deal with the human body; some of which are modifiable. Anatomical, hormonal, neuromuscular, biomechanical, and a previous history of an ACL injury are all subgroups of intrinsic risk factors for ACL injuries (Boden, et al., 2000; Griffin, et al., 2000; T.E. Hewett, et al., 2006; Ireland, 1999).

Anatomical risk factors include smaller intercondylar notch, knee hyperextension/genu recurvatum, greater Q-angle, greater navicular drop, greater joint laxity, and excessive flexibility or a debilitating lack of flexibility (T.E. Hewett, et al., 2006). These anatomical risk factors, as well as hormonal factors, cannot be easily controlled or modified. Some studies suggest that hormone levels during a female’s menstrual cycle
correlate with ACL injuries in female athletes. However, the literature is inconclusive; therefore definitive conclusions cannot be made at this time (Boden, et al., 2000; Wojtys, Huston, Lindenfeld, Hewett, & Greenfield, 1998). A prior history of an ACL injury is another risk factor that cannot be controlled.

It is important to know what causes ACL injuries, but if the risk factors are not easily modifiable, the data is of little clinical significance. For this reason, a significant amount of research focus has been on neuromuscular and biomechanical risk factors as these factors have been shown to be modifiable (J.D. Chappell & Limpisvasti, 2008; Chimera, Swanik, Swanik, & Straub, 2004). Some small and limited studies have even demonstrated some biomechanical and neuromuscular programs have even reduced injury rates (T. E. Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Mandelbaum, et al., 2005). However, there are other studies that have shown that intervention programs have not reduced injury rates, indicating a continuing need in research to determine what specific components make an effective intervention program (Pfeiffer, Shea, Roberts, Grandstrand, & Bond, 2006; Soderman K, 2000). Biomechanical and neuromuscular risk factors are challenging to differentiate and usually are discussed together as it is believed that one set of risk factors influences the other.

**Biomechanical Risk Factors**

Biomechanical factors suggested as ACL injury risk factors include a lesser hip flexion angle, lesser knee flexion angle, greater knee valgus angle, and greater vertical ground reaction forces when landing from a jump or cutting. In general, since females have a higher rate of ACL tears compared to males, sex differences in biomechanical and neuromuscular characteristics are believed to predisposing factors to ACL injuries. For instance, females have demonstrated a lesser hip flexion angle during athletic tasks compared to males (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007; McLean,
Lipfert, & van den Bogert, 2004). Landry et al. (2007) found that during a jump-stop task, males and females had similar hip flexion angles when taking off, but females landed with 48° of flexion and males landed with 56° of flexion. This same study found that females demonstrated a lesser hip flexion compared to males during a side-cut.

Females have also demonstrated lesser knee flexion angles in multiple studies; therefore it is believe that decreased knee flexion is a predisposing factor for ACL injuries (Decker, Torry, Noonan, Riviere, & Sterett, 2002; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Sigward, et al., 2008). A lesser knee flexion angle is associated with a significantly greater force placed on the ACL (Li, et al., 2004). Li et al. (2004) specifically found the force placed on the ACL was greatest during full extension and 30 degrees of knee flexion, but the force significantly decreased at 60 degrees of flexion and higher angles. A lesser knee flexion angle also increases the patella tendon-tibia shaft angle thus increasing the anterior shear force at the proximal end of the tibia produced by quadriceps contraction (Yu & Garrett, 2007). Theoretically, these extra forces placed on the ACL during lesser knee flexion angles could potentially predispose the ACL to injury.

Athletic maneuvers involve joint movements in the sagittal, frontal, and transverse planes. Knee valgus, which consists of a combination of femoral adduction and internal rotation, along with tibial abduction and external rotation, is a frontal plane movement that has also been linked to ACL injuries (Claiborne, et al., 2006; T.E. Hewett, et al., 2006). This is not surprising as previously stated, the load in the knee can be increased 6 times just by increasing knee valgus 5 degrees from neutral alignment (Bendijaballah, et al., 1997). Lateral femoral and tibial bone bruises associated with ACL injuries signify that lateral compression occurs when the medial knee joint opens up, indicating a valgus moment occurs during ACL injuries (Quatman & Hewett, 2009). Hewett et al. (2005) found that knee abduction had 78% sensitivity and 73% specificity in predicting ACL injury. Females have
demonstrated greater knee valgus in multiple studies (Malinzak, et al., 2001; McLean, et al., 2004; Russell, et al., 2006). All of these studies indicate why it is significant to determine what components contribute to knee valgus and in turn how to better prevent this motion during functional activity.

Ground reaction forces have also been considered to influence the risk of ACL injury due to an increase in ACL loading. Yu & Garrett (2007) demonstrated that greater posterior ground reaction forces increased ACL loading. Hewett et al. (2005) prospectively demonstrated that females who went on to tear their ACLs had significantly greater peak vertical ground reaction forces compared to those who did not.

In conclusion many biomechanical factors including lesser hip and knee flexion angles, along with greater knee valgus angles and ground reaction forces, have been considered as risk factors for ACL injuries. Hypothetically, correcting these biomechanical factors may potentially decrease the injury rate, but in order correct lower extremity kinematics neuromuscular factors will need to be considered as well.

**Neuromuscular Risk Factors**

Neuromuscular risk factors associated with ACL injuries incorporate muscle strength, contraction amplitudes, activation patterns, proprioception, and fatigue. Theoretically all of these components can contribute to poor biomechanics, including knee valgus. A considerable amount of research has been conducted to determine the effects of muscle strength and activation on knee biomechanics and how they, in turn, affect the ACL injuries.

**Muscle Strength**

There are multiple muscles that affect the hip and knee joint, along with the biomechanical risk factors associated with these respective joints. The quadriceps and hamstrings mainly control movements in the sagittal plane such as hip and knee flexion and extension. However both of these muscles have abductor and adductor moment arms, and
Palmieri-Smith et al. (2009) found that due to these moment arms, the quadriceps and hamstrings can provide dynamic frontal-plane knee stability. This indicates that a weakness or inadequate activation patterns or amplitudes of the quadriceps and hamstrings could lead to excessive knee valgus.

A greater quadriceps to hamstring strength ratio has been suggested as an ACL injury risk factor. Quadriceps dominance, which is defined by Hewett et al. (2001) as “an imbalance between quadriceps and hamstring recruitment patterns in which the quadriceps are activated over the hamstrings in an attempt to stabilize the knee” is theorized to increase ATSF, leading to greater ACL loading. For this reason a lot of research has been done investigating the quadriceps and hamstrings. Quadriceps dominance can either be due to a difference in quadriceps and hamstring strength or a difference in activation amplitudes and patterns. In a study conducted by Ahmad et al. (2006), 123 recreational soccer players were split into four groups consisting of premenarchal girls, girls 2 or more years post-menarche, boys under the age of 13, and boys older than 14. It was found that mature females had a significantly greater quadriceps to hamstring ratio (2.06) compared to the other three groups. Ahmad et al. (2006) also found that mature girls had 44% greater quadriceps and a 27% greater hamstring strength compared to the premenarchal girls, while mature boys had a 148% greater quadriceps strength increase and a 179% hamstring strength compared to the boys under the age of thirteen. These findings indicate that through maturity, females do not develop the same strength gains as their male counterparts, suggesting that strength could play a role in ACL injury. These findings are supported by a case study conducted by Myer et al. (2009) which found that as their female subject grew in height and mass, her hip and knee strength did not increase proportionally. However, contradictions in the literature also exist. For example, Shultz et al. (2009) studied hip and knee movement in the sagittal plane and did not find any significant correlations in muscle strength and knee kinematics when subjects performed a drop-jump task. Bennett
et al. (2008) performed a study investigating the correlation of quadriceps and hamstring strength to ATSF, as well as the correlation of the quadriceps/hamstring ratio and anterior tibial shear force. None of the variables were found to predict anterior tibial shear force (Bennett, et al., 2008).

The gluteal muscles have also been studied in the literature, but not as frequently as the quadriceps and hamstring muscle groups. These muscle groups need to be examined more extensively due to the potential effect they have on knee movement in the frontal plane. The gluteus medius’ function is to abduct and rotate the hip. The anterior fibers of the gluteus medius medially rotate the hip and may assist with flexion, while the posterior fibers laterally rotate the hip and may assist with extension (Kendall, et al., 2005). The gluteus medius also supports the pelvis during single-leg stance and prevents contralateral hip drop and ipsilateral genu valgus (Russell, et al., 2006). If muscle strength does indeed affect kinematic factors of the lower extremity, then a weak gluteus medius would directly affect knee valgus potentially by permitting more hip adduction during movement, thus contributing to greater knee valgus. A weak gluteus medius could also affect knee valgus due to the rotational components of the muscle. The gluteus maximus’ function is to extend and laterally rotate the hip (Kendall, et al., 2005). Theoretically, a weak gluteus maximus would cause a decrease in hip external rotation strength, thus allowing the hip to internally rotate, which would contribute to knee valgus.

There have been a couple of studies that have examined the gluteal muscles and kinematics. Weakness of the hip abductors has been detected in those demonstrating greater knee valgus and has been associated with other knee injuries including patellofemoral pain syndrome (Ireland, Willson, Ballantyne, & Davis, 2003; Jacobs, et al., 2007). Jacobs et al. (2007) demonstrated that women had lower hip abductor peak torque and greater knee valgus compared to males. Claiborne et al. (2006) found that hip, knee, and rotational strength factors, specifically hip abduction, hip internal rotation, knee flexion,
and knee extension strength, accounted for 74.3% of the variance in knee movement in the frontal plane during a single leg squat, and reported a significant negative weak-to-moderate correlation between concentric hip abduction, knee flexion, and knee extension strength and knee valgus. However, Bell et al. (2008) found that those with medial knee displacement, which can be associated with knee valgus, had significantly greater hip extension and hip external rotation strength compared to the control group, and Lawrence et al. (2008) found that hip external rotator strength did not significantly affect knee movement in the frontal plane. These contradictions indicate a continued need for research in this area. However, Bell et al. (2008) concluded that hip muscle strength does not influence lower extremity kinematics and instead suggested that muscle activation amplitudes and patterns could explain the differences in lower extremity kinematics.

Muscle Activation Amplitudes

A current conjecture is that people may have the strength to dynamically control their extremity, but lack the appropriate level of muscle activation. With this said, dynamic tasks, such as landing, do not require maximal muscular force production, so peak strength measures may not predict kinematics as much as once thought. Instead, the level of muscle activation may be a more important determinant of lower extremity kinematics (D. R. Bell, et al., 2008). Improper activation patterns and amplitudes such as an imbalance between the lateral and medial musculature, pre-activation of protective muscle groups, and fatigue could affect lower extremity biomechanical factors that predispose individuals to ACL injuries.

With this concept in mind, some of the current research has been focused on quadriceps and hamstring activation patterns and amplitudes for similar reasons as why studies focus on quadriceps and hamstring strength (Malinzak, et al., 2001; Palmieri-Smith, et al., 2009; Shultz, et al., 2009). Earlier studies have indicated that a quadriceps
contraction can apply an anterior shear force on the tibia through the patellar tendon, so if there is an imbalance in quadriceps and hamstring activation, with the quadriceps activating a high level, the ACL can be placed under greater stress compared to when there is equal activation among the muscle groups (Renstrom P., 1986). Shultz et al. (2009) found that even though a greater quadriceps activation amplitude was a significant predictor of greater anterior shear forces, quadriceps and hamstring activation patterns were not significant predictors of knee or hip movement in the sagittal plane. Palmieri-Smith et al. (2009) found that greater preparatory vastus medialis activity was associated with lesser knee valgus, while greater knee valgus was associated with greater preparatory activation of the vastus lateralis and lateral hamstring, indicating that activation patterns can affect knee kinematics.

Malinzak et al. (2001) demonstrated that females had greater quadriceps activation and lesser hamstring activation compared to their male counterparts during athletic tasks such as running, side-cutting, and cross-cutting. Hanson et al. (2008) specifically found that females had a greater vastus lateralis and quadriceps to hamstring ratio activation amplitudes during the preparatory and loading phases of side-step cutting. Zebis et al. (2009) performed a study that used EMG to test 55 elite female soccer or handball player’s pre-activity levels of the vastus lateralis, vastus medialis, rectus femoris, semitendinosus, and biceps femoris during a side-cut. The five athletes that went on to tear their ACLs demonstrated reduced pre-activity levels of the semitendinosus and increased pre-activity of vastus lateralis compared to the uninjured cohort. All of these studies indicate that activation patterns and amplitude could affect lower extremity kinematics.

Some studies have also investigated gluteal muscle activation amplitudes and patterns, which is a vital area to research due to the possible effects the gluteal muscles could have on knee movement in the frontal plane. Zazulak et al. (2005) did not find a significant gender difference in gluteus medius activation. Russell et al. (2006) also found that gluteus medius activation did not differ between genders during a single-leg drop jump
task, but females displayed greater knee valgus. The authors concluded that it is possible the timing of the activation is more important than the level of activation. Contradicting the previous studies, Hart et al. (2007) and Hanson et al. (2008) found a significant difference in the gluteus medius activity between genders. Unfortunately both studies did not examine knee kinematics as the previous studies did, thus it is unclear how this difference in gluteal activity influenced knee valgus.

The gluteus maximus has also been investigated in the literature. Hollman et al. (2009) found that gluteus maximus activation was negatively correlated with knee valgus and that it accounted for 20 percent of the variance in knee valgus during a step-down task. Another study found that females activated their gluteus maximus more than their male counterparts post-contact during a single-leg landing task (Zazulak, et al., 2005). Unfortunately kinematics were not measured in this study, again preventing the authors from identifying the influence of gluteal activity on knee valgus. The overall lack of research and the contradiction of the available research signifies the importance of continued research in this area. Inclusion of the gluteus maximus and medius research may aid in further understanding of the neuromuscular patterns that relate to the biomechanical risk factors associated with ACL injuries, including knee valgus.

Most studies focus their inquiry solely on strength or activation patterns and amplitudes; however, the relationship between muscle strength and activation has not been a primary interest in research. This relationship is important to understand because if muscle activation does indeed affect hip and knee kinematics, it’s essential to understand what causes differences in activation patterns and amplitudes. One thought is that decreased muscle strength requires greater activation to control the lower extremity, thus potentially causing the muscle to fatigue faster. Muscle fatigue can contribute to poor lower extremity biomechanics, which in turn could contribute to injury (J. D. Chappell, et al., 2005). One study that examined muscle strength and activation together found that thigh muscle
strength explains a portion of the variance in quadriceps and hamstring activation amplitudes (Shultz, et al., 2009).

It is important to understand the relationship between muscle strength and activation and how they affect lower extremity kinematics. By knowing these relationships, better injury prevention programs can be created. With this said, gaps in the current literature include the effect of hip musculature strength on gluteal muscle activation amplitudes. How these factors affect or predict ACL risk factors, including knee valgus, is also not clearly understand. In conclusion there is contradicting literature that suggests gluteal strength and activation influence ACL injury risk factors. This contradicting literature may suggest that it is a combination of these two factors that influence ACL injury risk factors instead of the factors in isolation. It is this theory that will be tested in this study.

**Jump Landing Task**

Landing from a jump has been indicated as one of the many mechanisms of injury for ACL tears (Griffin, et al., 2000). This specific task was the culprit in 6 of the 49 ACL tears investigated by Arendt et al. (1999). For this reason, this task has been used frequently in research studies (Barber-Westin SD, 2006; T. E. Hewett, et al., 2005; Joseph, et al., 2008; Shultz, et al., 2009). This task also involves rapid deceleration and acceleration, which also cause ACL injuries. Some studies position the box that the subjects jump off at a distance equal to 50 percent of each respective subject’s height (Beutler, et al., 2009; Padua, et al., 2009). By having the box at a distance that is 50 percent of the subject’s height, the task is made more challenging compared to the box being placed directly in front of the landing destination, and could potentially require heightened neuromuscular control. The distraction of having a subject jump for maximum height after landing from the initial jump in theory prevents a subject from focusing on the initial landing. In other words, it
stimulates game-like conditions. For these reasons, this specific jump landing task will be used for this study.

**Electromyography**

EMG is often used in the literature to assess muscle activation patterns and amplitudes of different hip and knee musculature (Ayotte, Stetts, Keenan, & Greenswa, 2007; Blackburn & Padua, 2009; Boudreau, et al., 2009; Landry, et al., 2007; Palmieri-Smith, et al., 2009; Shields, et al., 2005; Shultz, et al., 2009). EMG provides an indication of the neural drive sent from the central nervous system to the muscle. An amplifier magnifies the muscle action potentials and smoothes out ambient noise (Pease, Lew, & Johnson, 2007). It is imperative to understand that EMG is not a measure of muscle strength, but a measure of the electricity activity in a muscle.

When performing EMG studies it is important to place the electrodes parallel to the muscle fibers and in the middle of the muscle belly so that they are not close to the neuromuscular junction (Ayotte, et al., 2007). It is also critical to clean and abrade the skin in order to remove any residue of oils, lotions, perfumes, and dead skin. Lotions and perfumes are poor conductors of electricity and dead skin offers high impedance (Pease, et al., 2007).

All EMG recordings require three electrodes that include a reference and two active electrodes. In this study, bipolar electrodes will be used, which contain two electrodes at the recording site and the reference electrode at an electrically neutral site, the tibial tuberosity. Electrodes are the sensors that are responsible for detecting electrical action potentials generated in the muscle. An action potential is defined “as a membrane response in a nerve or muscle after reaching excitation threshold” (Pease, et al., 2007). Surface electrodes are the easiest and most cost-effective electrodes to use, but they lack specificity, are limited to large surface muscles, and move with the skin.
Leads or cables attach the electrodes to an amplifier. The amplifier is probably the most important component of the EMG system as it magnifies the potential difference between the active and reference inputs. An analog to digital converter is then used to measure the EMG signal at regular time intervals which are then plotted and connected by a line (Pease, et al., 2007).

The Nyquist theorem states that EMG signals should be sampled at a rate that is at least twice the frequency of its highest harmonic order (Merletti & Parker, 2004). Filters are then used to attenuate noise. A high-pass filter is a lower frequency filter that accepts data above a certain frequency, while a low-pass filter is a higher frequency filter that accepts data below a certain frequency. A high-pass filter is usually around 10-20 Hz because harmonics of unwanted artifacts in surface electrodes occur around the 0-20 Hz range. A low-pass filter is usually around 400-450 Hz (Merletti & Parker, 2004). A bandpass filter has a low and high setting to smooth out “noise”. Decreasing the high-frequency and increasing the low-frequency, thus reducing the bandwidth, affects the EMG signal indicating the importance of understanding the EMG waveform being collected. A notch filter is used to smooth out specific amplitudes and is typically used around 50-60 Hz. At this specific amplitude power-lines operate which is why it is important filter out this signal, as it could compromise the sample.

**Clinical Significance**

Injury prevention is a very important role of a clinician, as it not only saves time lost by the athlete, but it also saves money. As stated previously, a significant amount of money is spent each year in repairing and rehabilitating ACL injuries. ACL research has come a long way in offering a better understanding of ACL injuries and associated risk factors, but there are areas that still need to be investigated. The effect of hip strength on biomechanical risk factors has been studied but the results and conclusions are
inconclusive. There is also little research incorporating how muscle strength relates to muscle activation amplitudes. This study will add to the literature on how hip strength affects lower extremity kinematics, specifically knee valgus and factors associated with knee valgus. More insight on how hip strength affects gluteus maximus and gluteus medius activation amplitudes will be also obtained, along with how these two factors predict knee valgus and the associated movements. The outcomes of this study could also contribute to current injury prevention programs; specifically if gluteus maximus and medius strengthening or neuromuscular training could reduce knee valgus.
Chapter 3

Methodology

Experimental Design

This cross-sectional investigation utilized a casual-comparative design to evaluate the influence of gluteal strength and EMG amplitudes on peak knee valgus, hip adduction, and hip internal rotation angles. Hip extension, abduction, and external rotation strength were always measured first via a hand-held dynamometer, with the order of muscle testing being randomized. Lower extremity kinematics were then measured using an electromagnetic motion capture system as the subjects performed a double-leg jump landing task. Gluteus medius and maximus EMG data were also sampled during each of these tasks. All data was sampled from the subject’s dominant limb, which was defined as the leg that would be used to kick a ball for maximal distance.

Subjects

A sample of 82 physically active, healthy subjects between the ages of 18-30 years was recruited to participate in this study. Demographics of the subjects can be seen in Table 3.1. Physically active was defined as participation in at least 30 minutes of physical activity 3 times per week. Subjects were excluded from participation if they had a history of back or lower extremity surgery, ACL injury, back or lower extremity injury in the 6 months prior to data collection that prevented them from carrying out activities of everyday life, or
chronic or neurological disorders. Upon reporting to the Neuromuscular Research
Laboratory (NMRL), subjects read and sign an informed consent document approved by the
Institutional Review Board at the University of North Carolina at Chapel Hill. Anthropometric
measurements consisting of the subject’s height (cm) and mass (kg) were then recorded.

Measurement and Instrumentation

A hand-held dynamometer (Chatillon CSD 300, Amteck, Inc., Largo, FL) was used to
measure peak muscle force during maximum voluntary isometric contractions (MVICs) of
the hip abductors, extensors, and external rotators. All strength data was collected by the
same researcher (MFN) to enhance reliability. Table 3.2 provides the intraclass correlation
coefficients (ICC) and standard errors of measurements (SEMs) for these strength
measurements.

An electromagnetic motion capture system (MotionStar, Ascension Technology
Corp., Burlington, VT) was used to assess kinematic data including peak knee valgus, hip
adduction, and hip internal rotation angles. All kinematic data was collected at 120Hz. A
non-conductive force plate (Model 4060-NC; Bertec Corporation, Columbus, OH) was used
to capture ground reaction forces simultaneously at 1,200Hz.

Neural activity of the gluteus maximus and gluteus medius was measured during the
MVIC and kinematic data collection using pre-amplified/active surface EMG electrodes,
which have an interelectrode distance of 10mm. The signals were amplified by a factor of
10,000 (DelSys Bagnoli-8, DelSys Inc., Boston, MA). The signal amplifier features a
common-mode rejection ratio of 80dB and an input impedance greater than 10^15 ohms/0.2
pF. EMG data was sampled at 1,200Hz. The Motion Monitor motion capture software
(Innovation Sports Training, Chicago, IL) was used for kinematic model calibration and to
synchronize collection of the kinematic, kinetic, and EMG data.
**Procedures**

*Isometric Muscle Strength Testing*

The MVICs for each subject’s dominant hip abductor, extensor, and external rotator were measured in a randomized order via a hand-held dynamometer. Three trials were conducted for each muscle group and measured by the same investigator (MFN) to improve reliability. Subjects were instructed to maximally contract for 5 seconds as the investigator resisted the motion, thus maintaining an isometric contraction. All subjects received verbal encouragement. Peak force measured using the dynamometer was multiplied by segment length to calculate peak torque. Results were then normalized to the product of body weight and height (Nm).

*Hip abduction strength*

Hip abduction strength was measured with the subject side-lying on the non-dominant side. The non-dominant leg was slightly flexed at the knee and hip in order to help the subject remained balanced. The subject was asked to abduct the dominant hip without flexing, extending, or rotating the hip and to keep the ipsilateral knee fully extended. The dynamometer was placed on the lateral aspect of the subject’s thigh just proximal to the lateral joint line of the knee. The subject was asked to push into the dynamometer with maximal effort while the investigator resisted motion (Hislop, Mongomery, Connelly, & Daniels, 1995). Figure 3.1 demonstrates the subject and dynamometer positioning used for this measurement.

*Hip extension strength*

Hip extension strength was measured with the subject lying prone with the dominant knee flexed to 90°. The subject was instructed to lift the dominant foot toward the ceiling, while keeping the ipsilateral knee flexed to 90 degrees and not rotating the hips. This
position was selected in order to isolate the gluteus maximus muscle and minimize contributions from the hamstrings. The investigator placed the dynamometer immediately superior to the knee, over the posterior thigh. The subject was asked to push into the dynamometer with maximal effort while the investigator resisted this motion (Hislop, et al., 1995). Figure 3.2 demonstrates the subject and dynamometer positioning used for this measurement.

**Hip external rotation strength**

Hip external rotation strength was measured with the subject prone, the knee flexed to 90 degrees, and the hip in a neutral position. The subject was instructed to maximally externally rotate the dominate hip against the dynamometer positioned over the medial aspect of the shank, just proximal to the medial malleolus. The investigator applied a force over the medial aspect of the ankle in the lateral direction (Hislop, et al., 1995). Figure 3.3 demonstrates the subject and dynamometer positioning used for this measurement.

**Electromyography**

EMG electrodes were placed over the gluteus maximus and gluteus medius muscle bellies parallel with the muscle fibers (Ayotte, et al., 2007). All sites for electrode placement were prepared by shaving, abrading, and cleaning the skin with alcohol pads in order to reduce impedance.

Electrode placement on the gluteus maximus was located at the proximal 1/3 of the distance between the greater trochanter of the femur and the spinous process of the S2 vertebrae (Rainoldi, Melchiorri, & Caruso, 2004). The electrodes placed on the gluteus medius were positioned so that they were located at the distal 1/3 of the distance between the iliac crest and greater trochanter of the femur (Rainoldi, et al., 2004). Both electrode placements were confirmed with an isometric manual muscle test and checked for cross talk by viewing the EMG monitor and making sure the EMG signals from the two muscles did not
have the same shape. The reference electrode was placed on the tibial tuberosity. Intraclass correlation coefficients using this technique for EMG normalization for the gluteus medius and maximus have been reported to be 0.98 and 0.95 respectively (Norcross, Blackburn, & Goerger, 2009).

**Jump Landing Task**

Subjects performed a double-leg jump landing from a 30 cm box positioned a distance equal to 50% of their height away from the force plate. Subjects took three practice jumps, but were permitted to take as many practice trials needed in order to become comfortable with the task. Subjects then performed 5 successful trials with 30 seconds between each trial in order to reduce the likelihood of fatigue. Successful trials required the subjects to land with the dominant foot completely on the force plate and the non-dominant foot completely off the force plate, and to immediately jump vertically for maximum height (Padua, et al., 2009) Unsuccessful trials were repeated until 5 successful trials were obtained.

Hip and knee joint kinematics, ground reaction forces, and gluteal EMG measures were sampled simultaneously during the jump landing task via The Motion Monitor software. Global and segment axis systems were established using the right-hand coordinate system where positive X was anterior, positive Y was medial, and positive Z was superior. Three electromagnetic sensors were placed on the pelvis, thigh, and shank with double-sided tape and secured with pre-wrap and tape. A segment-linkage model of the dominant lower extremity was then derived by digitizing landmarks around the hip, knee, and ankle joint centers using a stylus. The location of the hip joint center was estimated using the Bell Method. The knee and ankle joint centers were defined as the midpoint between the digitized lateral and medial femoral condyles and lateral and medial malleoli, respectively.
Data Sampling and Reduction

Kinematic data was sampled at 120 Hz and low pass filtered at 12 Hz (4th order zero-phase-lag Butterworth), while kinetic and EMG data were sampled at 1200 Hz. The kinematic data was time-synchronized to the EMG and kinetic data and re-sampled to 1,200 Hz. Knee and hip kinematic angles were calculated using Euler angles in a Y X’ Z’’ rotation sequence. Euler angles were defined as flexion and extension occurring about the Y-axis, adduction and abduction occurring about the X-axis, and internal and external rotation occurring about the Z-axis. Motion about the knee was defined as the shank relative to the thigh and about the hip as the thigh relative to the sacrum. EMG data was corrected for DC bias and band-pass (20-350 Hz) and notch (59.5-60.5 Hz) filtered (4th order zero-phase-lag Butterworth) using custom software (LabVIEW, National Instruments, Austin, TX).

Peak hip adduction, hip internal rotation, and knee valgus angles, along with the peak EMG amplitudes for the gluteus medius and maximus were calculated during the loading phase of the landing task. If a subject did not reach hip adduction, hip internal rotation, or knee valgus during the task, the value closest to these positions was recorded. The loading phase was defined as the time interval from initial ground contact to maximum knee flexion (Blackburn & Padua, 2009). Initial ground contact was defined as the instant at which the vertical ground reaction force exceeded 10N. The peak amplitudes during the MVIC trials were also analyzed. The averages across the 3 trials for each respective muscle were used to normalize the peak EMG amplitudes of the respective muscles during the jump landing task. The normalized peak EMG amplitudes were reported as a percentage for each subject. During data collection when conducting the MVIC trials for hip external rotation there was not always a definitive spike in EMG amplitude when the muscle was activated, unlike the hip extension and abduction trials. Because of this observation,
we normalized all gluteus maximus EMG amplitudes to the hip extension MVIC EMG amplitudes.

**Statistical Analysis**

The 82 subjects were arranged into tertiles based on hip strength (n = 27 for the high and low tertiles; n = 28 for the middle tertile). Gluteus medius and maximus EMG amplitudes, peak knee valgus, peak hip adduction, and peak hip internal rotation angles were compared between the highest and lowest tertiles for each strength test via independent-samples t-tests. The middle subjects were excluded from these analyses in an effort to create two groups with disparate strength values.

Multiple linear regression analyses were used to evaluate the relationships between peak kinematic values during the jump landing task and the linear combinations of muscle strength and EMG amplitude. Specifically, separate multiple linear regression models were used to evaluate

1) the relationship between peak hip adduction and the linear combination of hip abduction strength and gluteus medius EMG amplitude

2) the relationship between peak hip internal rotation and the linear combination of hip external rotation strength and gluteus maximus EMG amplitude

3) the relationship between peak hip internal rotation and the linear combination of hip extension strength and gluteus maximus EMG amplitude

4) the relationship between peak knee valgus and the linear combination of hip abduction strength and gluteus medius EMG amplitude
5) the relationship between peak knee valgus and the linear combination of hip external rotation strength and gluteus maximus EMG amplitude

6) the relationship between peak knee valgus and the linear combination of hip extension strength and gluteus maximus EMG amplitude.

All data was analyzed using PASW 18.0 (Chicago, IL) statistical software with statistical significance established a priori as $\alpha \leq 0.05$. Table 3.2 provides descriptions of each of the statistical analyses noted above.
Figure 3.1: Subject and dynamometer positioning for hip abduction isometric strength testing
Figure 3.2: Subject and dynamometer positioning for hip extension isometric strength testing
Figure 3.3: Subject and dynamometer positioning for hip external rotation isometric strength testing
### Table 3.1. Subject Demographics

<table>
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<th>Males (n=41)</th>
<th>Females (n=41)</th>
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<tr>
<td>Age (years)</td>
<td>21.1 ± 2.1</td>
<td>20.8 ± 2.7</td>
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<tr>
<td>Height (cm)</td>
<td>181.33 ± 6.40</td>
<td>166.51 ± 6.15</td>
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<tr>
<td>Mass (kg)</td>
<td>77.61 ± 13.20</td>
<td>61.19 ± 9.17</td>
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Table 3.2: Intraclass correlation coefficients (ICC) and standard errors of the mean (SEM)

<table>
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<tr>
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<th>ICC (2,1)</th>
<th>SEM (N*m)</th>
<th>SEM as a % of the mean value</th>
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</thead>
<tbody>
<tr>
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<td>9.49</td>
<td>7.8</td>
</tr>
<tr>
<td>Hip Abduction</td>
<td>0.97</td>
<td>7.70</td>
<td>6.3</td>
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<td>Hip External Rotation</td>
<td>0.96</td>
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<td>6.8</td>
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</table>
### Table 3.3. Statistical Analyses

<table>
<thead>
<tr>
<th>Question</th>
<th>Description</th>
<th>Data Source</th>
<th>Comparison</th>
<th>Method</th>
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</table>
| 1                                                                       | Do hip and knee kinematics during the loading phase of a double-leg jump landing task differ between groups displaying higher and lower hip abduction, extension, and external rotation strength? | DV:  
- peak hip adduction angle  
- peak hip internal rotation angle  
- peak knee valgus angle | Kinematic measures (peak knee valgus, hip adduction, and internal rotation) in subjects with higher strength measures to subjects with lower strength | Six Independent T-Tests |
Chapter 4

Results

Independent t-tests signify that the high and low strength tertile groups were significantly different for hip external rotation ($t_{52}=16.916$, $p<0.001$), hip abduction ($t_{52}=12.013$, $p<0.001$), and hip extension ($t_{52}=13.064$, $p<0.001$), indicating that there was a significant difference in the “strong” group and “weak” groups. Means and standard deviations for each strength comparison are detailed in Table 4.1. It is important to note that the strong groups were primarily composed of males and the weak groups were primarily composed of females. The demographics for each tertile are detailed in Table 4.2. The results for the independent t-tests used to evaluate differences in lower extremity kinematics and peak EMG amplitudes between high and low hip strength groups are describe below; as well as the results for the regression analyses investigating if strength and peak EMG amplitudes of the gluteal muscles predict peak knee valgus angle and other kinematic factors associated with knee valgus including peak hip adduction and internal rotation.

Hip Strength and Kinematics

There were no significant differences in peak hip adduction, hip internal rotation, or knee valgus angles between those demonstrating high vs. low isometric hip abduction or extension strength, respectively. Some of these differences approached the alpha level set
at 0.05 thus we conducted secondary analyses to evaluate the observed power and
determine the number of subjects that would have been necessary to achieve a priori power
of 0.80. These analyses indicated generally a low observed power and that a much higher
number of subjects (80-90 subjects per group) would have been necessary to provide
adequate power. There were a significant differences in peak hip external rotation and knee
valgus angles between the high and low hip external rotation strength groups. Those with
greater peak isometric hip external rotation strength demonstrated lesser peak hip external
rotation \( t_{52}=2.033, p=0.0236 \) and knee valgus \( t_{52}=2.209, p=0.0158 \) compared to those
with lesser peak isometric hip external rotation strength. All of these results can be viewed
in Tables 4.3, 4.4, and 4.5

**Hip Strength and EMG Amplitude**

There was no significant difference in the gluteus medius EMG amplitudes during the
loading phase of a double-leg jump landing between those demonstrating high vs. low
isometric hip abduction strength. There were, however, significant differences in gluteus
maximus EMG amplitude between the hip extension strength groups \( t_{52}=-1.902, p=0.031 \)
and hip external rotation strength groups \( t_{52}=-1.749, p=0.043 \), with stronger individuals
demonstrating less gluteus maximus EMG activity. The results for the means and standard
deviations for all the independent t-tests run can be seen in Tables 4.3, 4.4, 4.5.

**Regressions**

Regression analyses indicated that isometric hip external rotation strength and
gluteus maximus peak EMG amplitudes significantly predicted 7.9% of the variance in peak
hip rotation angle \( r= 0.281, p=0.039. \) and 12.1% of the variance in peak knee valgus angle
\( r=0.348, p=0.006 \) during the loading phase of a double-leg jump landing task. Analysis of
the significant findings show that peak isometric hip rotation strength (standardized Beta-
weight = 0.284) explained a greater portion of the variance in peak hip motion in the
transverse plane compared to gluteal maximus activation amplitude (standardized Beta-weight = 0.024). Similarly, isometric hip rotation strength (standardized Beta-weight = 0.348) was a stronger predictor of peak knee valgus angle compared to gluteal maximus activation amplitude (standardized Beta-weight = 0.124). The results from these significant findings can be seen in Figures 4.1 and 4.2. No other regressions were found to be significant. Dependent and predictor variables for each regression can be found in Table 4.6. Descriptions for each regression analyzed can be viewed in Tables 4.7-4.12.
Table 4.1. Muscle strength normalized to body mass and length of the segment.

<table>
<thead>
<tr>
<th></th>
<th>High tertile (n=27)</th>
<th></th>
<th>Low tertile (n=27)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>Standard deviation</td>
<td>Mean</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>Hip external rotation strength (%BW*Ht)</td>
<td>0.0575*</td>
<td>0.0057</td>
<td>0.0363</td>
<td>0.0030</td>
</tr>
<tr>
<td>Hip abduction strength (%BW*Ht)</td>
<td>0.1317*</td>
<td>0.0235</td>
<td>0.0754</td>
<td>0.0065</td>
</tr>
<tr>
<td>Hip extension strength (%BW*Ht)</td>
<td>0.1272*</td>
<td>0.0170</td>
<td>0.0785</td>
<td>0.0092</td>
</tr>
</tbody>
</table>

*p<0.05
Table 4.2. Demographics of strength tertiles

<table>
<thead>
<tr>
<th></th>
<th>High tertile (n=27)</th>
<th>Low tertile (n=27)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Males</td>
<td>Females</td>
</tr>
<tr>
<td>Hip external rotation strength</td>
<td>22</td>
<td>5</td>
</tr>
<tr>
<td>Hip extension strength</td>
<td>19</td>
<td>8</td>
</tr>
<tr>
<td>Hip abduction strength</td>
<td>21</td>
<td>6</td>
</tr>
</tbody>
</table>
Table 4.3. Differences in kinematics and peak EMG amplitudes between high vs. low hip abduction strength

<table>
<thead>
<tr>
<th>Dependent Variable</th>
<th>High Strength Tertile (n=27)</th>
<th>Low Strength Tertile (n=27)</th>
<th>Observed Power</th>
<th>Subjects/group necessary for 0.80 priori power</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Hip Adduction Angle (º) †</td>
<td>1.98 ± 8.31</td>
<td>3.19 ± 6.77</td>
<td>0.161</td>
<td>481</td>
</tr>
<tr>
<td>Peak Knee Valgus Angle (º) ‡</td>
<td>-15.34 ± 9.00</td>
<td>-18.92 ± 9.34</td>
<td>0.372</td>
<td>90</td>
</tr>
<tr>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td>101.08 ± 117.15</td>
<td>155.10 ± 177.36</td>
<td>0.367</td>
<td>93</td>
</tr>
</tbody>
</table>

† Negative value indicates hip abduction and a positive value indicates hip adduction
‡ Negative value indicates knee valgus and a positive value indicates knee varus
*p<0.05
<table>
<thead>
<tr>
<th>Dependent Variable</th>
<th>High Strength Tertile (n=27)</th>
<th>Low Strength Tertile (n=27)</th>
<th>Observed Power</th>
<th>Subjects/group necessary for 0.80 priori power</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Hip Internal Rotation Angle (°) †</td>
<td>-3.21 ± 9.92*</td>
<td>-8.39 ± 8.75</td>
<td>_</td>
<td>_</td>
</tr>
<tr>
<td>Peak Knee Valgus Angle (°) ‡</td>
<td>-14.65 ± 11.28*</td>
<td>-21.17 ± 10.39</td>
<td>_</td>
<td>_</td>
</tr>
<tr>
<td>Peak Gluteus Maximus EMG Amplitude (%MVIC)</td>
<td>117.39 ± 63.20*</td>
<td>307.00 ± 559.75</td>
<td>_</td>
<td>_</td>
</tr>
</tbody>
</table>

† Negative value indicates hip external rotation and a positive value indicates hip internal rotation
‡ Negative value indicates knee valgus and a positive value indicates knee varus
*p<0.05
Table 4.5. Differences in kinematics and peak EMG amplitudes between high vs. low hip extension strength

<table>
<thead>
<tr>
<th>Dependent Variable</th>
<th>High Strength Tertile (n=27)</th>
<th>Low Strength Tertile (n=27)</th>
<th>Observed Power</th>
<th>Subjects/group necessary for 0.80 priori power</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Hip Internal Rotation Angle (°)†</td>
<td>-3.77 ± 8.95</td>
<td>-7.61 ± 9.04</td>
<td>0.427</td>
<td>69</td>
</tr>
<tr>
<td>Peak Knee Valgus Angle (°)‡</td>
<td>-15.46 ± 9.00</td>
<td>-19.27 ± 10.61</td>
<td>0.388</td>
<td>83</td>
</tr>
<tr>
<td>Peak Gluteus Maximus EMG Amplitude (%MVIC)</td>
<td>117.72 ± 50.08*</td>
<td>322.22 ± 556.49</td>
<td>—</td>
<td>—</td>
</tr>
</tbody>
</table>

†Negative value indicates hip external rotation and a positive value indicates hip internal rotation
‡Negative value indicates knee valgus and a positive value indicates knee varus
*p<0.05
<table>
<thead>
<tr>
<th>Dependent Variable</th>
<th>Predictor Variables</th>
<th>$R^2$</th>
<th>p value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Hip Rotation</td>
<td>Hip external rotation strength</td>
<td>0.079</td>
<td>0.039</td>
</tr>
<tr>
<td></td>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee Valgus</td>
<td>Hip external rotation strength</td>
<td>0.121</td>
<td>0.006</td>
</tr>
<tr>
<td></td>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak Hip Frontal Plane Position</td>
<td>Hip abduction strength</td>
<td>0.002</td>
<td>0.920</td>
</tr>
<tr>
<td></td>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee Valgus</td>
<td>Hip abduction strength</td>
<td>0.036</td>
<td>0.237</td>
</tr>
<tr>
<td></td>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak Hip Rotation</td>
<td>Hip extension strength</td>
<td>0.039</td>
<td>0.209</td>
</tr>
<tr>
<td></td>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee Valgus</td>
<td>Hip extension strength</td>
<td>0.056</td>
<td>0.102</td>
</tr>
<tr>
<td></td>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
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</tr>
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</table>
Table 4.7. Predictability of peak hip internal rotation based on hip external rotation strength and gluteus maximus activation

<table>
<thead>
<tr>
<th>Variable</th>
<th>Parameter Estimate</th>
<th>SE</th>
<th>Standardized Coefficient</th>
<th>t Value</th>
<th>p Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Hip Rotation</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Intercept</td>
<td>-18.248</td>
<td>5.262</td>
<td>-3.468</td>
<td>0.001</td>
<td></td>
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<tr>
<td>Hip external rotation strength</td>
<td>280.880</td>
<td>108.367</td>
<td>0.284</td>
<td>2.592</td>
<td>0.011</td>
</tr>
<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.001</td>
<td>0.003</td>
<td>0.024</td>
<td>0.222</td>
<td>0.825</td>
</tr>
<tr>
<td>Variable</td>
<td>Parameter Estimate</td>
<td>SE</td>
<td>Standardized Coefficient</td>
<td>t Value</td>
<td>p Value</td>
</tr>
<tr>
<td>----------</td>
<td>-------------------</td>
<td>-----</td>
<td>--------------------------</td>
<td>---------</td>
<td>---------</td>
</tr>
<tr>
<td>Knee Valgus</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Intercept</td>
<td>-35.703</td>
<td>5.750</td>
<td>-6.210</td>
<td>&lt; 0.001</td>
<td></td>
</tr>
<tr>
<td>Hip external rotation strength</td>
<td>383.977</td>
<td>118.408</td>
<td>0.348</td>
<td>3.243</td>
<td>0.002</td>
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<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.004</td>
<td>0.003</td>
<td>0.124</td>
<td>1.156</td>
<td>0.251</td>
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</table>
Table 4.9. Predictability of peak hip adduction based on hip abduction strength and gluteus medius activation

<table>
<thead>
<tr>
<th>Variable</th>
<th>Parameter Estimate</th>
<th>SE</th>
<th>Standardized Coefficient</th>
<th>t Value</th>
<th>p Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frontal Plane Hip Position</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Intercept</td>
<td>3.931</td>
<td>3.310</td>
<td>1.187</td>
<td>0.239</td>
<td></td>
</tr>
<tr>
<td>Hip abduction strength</td>
<td>-10.357</td>
<td>30.551</td>
<td>-0.038</td>
<td>-0.339</td>
<td>0.736</td>
</tr>
<tr>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td>0.001</td>
<td>0.006</td>
<td>0.025</td>
<td>0.223</td>
<td>0.824</td>
</tr>
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</table>
Table 4.10. Predictability of peak knee valgus based on hip abduction strength and gluteus medius activation

<table>
<thead>
<tr>
<th>Variable</th>
<th>Parameter Estimate</th>
<th>SE</th>
<th>Standardized Coefficient</th>
<th>t Value</th>
<th>p Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee Valgus</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Intercept</td>
<td>-24.707</td>
<td>4.626</td>
<td>-</td>
<td>-5.341</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Hip abduction strength</td>
<td>72.899</td>
<td>42.689</td>
<td>0.189</td>
<td>1.708</td>
<td>0.092</td>
</tr>
<tr>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td>0.001</td>
<td>0.009</td>
<td>0.018</td>
<td>0.161</td>
<td>0.873</td>
</tr>
</tbody>
</table>
Table 4.11. Predictability of peak hip internal rotation based on hip extension strength and gluteus maximus activation

<table>
<thead>
<tr>
<th>Variable</th>
<th>Parameter Estimate</th>
<th>SE</th>
<th>Standardized Coefficient</th>
<th>t Value</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip Rotation</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Intercept</td>
<td>-13.626</td>
<td>5.041</td>
<td>-0.202</td>
<td>-2.703</td>
<td>0.008</td>
</tr>
<tr>
<td>Hip extension strength</td>
<td>82.725</td>
<td>46.705</td>
<td>0.202</td>
<td>1.771</td>
<td>0.080</td>
</tr>
<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.001</td>
<td>0.003</td>
<td>0.025</td>
<td>0.219</td>
<td>0.827</td>
</tr>
</tbody>
</table>
Table 4.12. Predictability of peak knee valgus based on hip extension strength and gluteus maximus activation

<table>
<thead>
<tr>
<th>Variable</th>
<th>Parameter Estimate</th>
<th>SE</th>
<th>Standardized Coefficient</th>
<th>t Value</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee Valgus</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Intercept</td>
<td>-28.871</td>
<td>5.587</td>
<td>-5.167</td>
<td>&lt;0.001</td>
<td></td>
</tr>
<tr>
<td>Hip extension strength</td>
<td>108.185</td>
<td>51.765</td>
<td>0.236</td>
<td>2.090</td>
<td>0.040</td>
</tr>
<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.004</td>
<td>0.003</td>
<td>0.122</td>
<td>1.079</td>
<td>0.284</td>
</tr>
</tbody>
</table>
Figure 4.1. Correlation between hip strength and hip rotation during a jump landing task

*Positive values indicate internal rotation and negative values indicate external rotation*
Figure 4.2. Correlation between hip strength and knee valgus during a jump landing task

*Positive values indicate knee varus and negative values indicate knee valgus
Chapter 5

Discussion

The primary findings of this study include the following: individuals with greater peak isometric hip external rotation strength demonstrated lesser peak hip external rotation and knee valgus angles, and those with stronger hip extensors and external rotators demonstrated a lesser peak gluteus maximus amplitudes than those with weaker hip extensors and external rotators respectively. Additionally, isometric hip external rotation strength and gluteus maximus EMG amplitude explained a significant, though limited, amount of variance in peak hip rotation and knee valgus angles during the loading phase of a double-leg jump landing task. Furthermore, hip abduction and extension strength, and gluteus medius EMG activity did not influence peak hip or knee kinematics.

Some of these findings agree with previous literature, while others differ from the already controversial literature on the influences of gluteal muscle strength and activation on hip and knee biomechanics. While a number of previous investigations evaluated differences in these variables between and within genders, the goal of this study was to evaluate differences between stronger and weaker individuals. However, it is important to note that the stronger tertiles for each strength measure were composed primarily of males while the weaker tertiles were composed primarily of females. The implications of this sex stratification will be discussed later in this chapter.
Individuals with greater isometric hip external rotation strength demonstrated lesser peak knee valgus angles compared to weaker individuals. This difference suggests that subjects with stronger hip external rotators possess a better capacity for resisting knee movement in the frontal plane. Similarly, Willson et al. (2006) found that greater hip external rotation strength was associated a lesser 2D frontal plane knee projection angle ($r = 0.40$) during a single-leg squat in males and females. Conversely, Lawrence et al. (2008) found no significant difference in knee valgus angles during a single-leg drop landing task between females demonstrating greater vs. lesser hip external rotation strength. The contradiction in the results between the former study and ours brings the influence of hip strength on kinematics into question. One noticeable difference between the two studies is the populations tested. When we limited our statistical analysis to stronger vs. weaker females there was not a significant difference in peak knee valgus angle between those demonstrating stronger and weaker hip rotation strength ($t=0.289, p=0.387$). The results for the means and standard deviations for all the secondary analyses can be found in Table 5.1. This finding agrees with Lawrence et al. (2008) and indicates that the difference in kinematics maybe due to sex differences, not necessarily strength. This concept will be discussed in more detail later in this chapter.

Our study also found that individuals with stronger hip external rotators displayed smaller peak hip external rotation angles compared to weaker individuals. Hip internal rotation is one component of knee valgus (Claiborne, et al., 2006; Hewett, et al., 2006). Therefore, greater hip internal rotation would seemingly be associated greater knee valgus. As such, we hypothesized that individuals with greater hip external rotation strength would demonstrate less hip internal rotation and, therefore, less knee valgus. However, our data indicates that individuals with greater hip external rotation strength demonstrated less knee valgus, but also demonstrated less peak external rotation (i.e. a more internally rotated hip) compared to those with weaker external rotators. Both the stronger and weaker hip external
rotation groups demonstrated a peak hip transverse plane angles representing hip external rotation. This finding indicates that neither group’s transverse plane motion reached internal rotation as defined by our kinematic conventions, but that the weaker group demonstrated greater hip external rotation angles. Peak kinematic values provide a limited indication of joint motion; joint displacements provide additional information, as they represent the amount of motion occurring between initial ground contact and the peak angular value. When we investigated the difference in displacements between the stronger and weaker external rotation groups, there was a significant difference (t=-2.185, p=0.017), with the stronger group demonstrating less displacement (6.79º ± 1.16) compared to the weaker group (10.18º ± 1.02). This indicates that the weaker group started in a greater externally rotated position and moved through a greater range of motion toward hip internal rotation. With the subject’s foot planted, the hip movement toward internal rotation results in greater peak knee valgus. It is possible that the weaker group landed in a toe-out position resulting in a greater starting hip and tibial external rotation position. In this hypothetical situation, any femoral internal rotation would increase the knee valgus position, indicating that foot position could influence kinematics. Future research needs to be conducted to determine the exact relationship between these two factors, along with the role of displacements on kinematics. There were no other significant differences in kinematics between the strong and weak groups, but some of these comparisons approached statistical significance. The group displaying greater hip abduction strength demonstrated lesser knee valgus angles compared to the weaker group (p=0.089). Other studies have found a significant difference in peak knee valgus angles between those demonstrating stronger and weaker hip abduction strength (Ireland, et al., 2003; Jacobs, et al., 2007). The group displaying greater hip extension strength demonstrated lesser peak hip external rotation (p=0.061) and knee valgus angles (p=0.080) compared to the weaker group. Similar to our findings regarding
hip external rotation strength, individuals with stronger hip extensors demonstrated a less externally rotated hip. These individuals did not, however, display different hip internal rotation displacements (strong= 8.34° ± 5.72, weak= 8.55° ± 5.26). As previously stated more research needs to be done investigate displacements and their role in predicted kinematics that are believed to increase injury rates.

Bell et al. (2008) found that individuals with excessive medial knee displacement, which is associated with knee valgus, during an overhead squat had significantly greater hip extension and hip external rotation strength compared to those subjects displaying a neutral frontal plane angle. These authors speculated that possessing adequate strength to control the lower extremity during dynamic tasks does not necessitate the appropriate level of muscle activation. With this said, dynamic tasks such as landing do not require maximal muscular force production, thus peak strength measures may not adequately predict kinematics. Instead, the level of muscle activation may be a more important determinant of lower extremity motion (Bell, et al., 2008). Our study found that individuals with greater hip extension and external rotation strength demonstrated lesser gluteus maximus EMG amplitudes compared to weaker individuals. Therefore, a weaker muscle requires greater activation to control the lower extremity compared to a stronger muscle. Theoretically, if a muscle is weak, greater muscle activity could be used to compensate for the mechanical weakness, and if an increase in neural drive is needed to control the lower extremity, the likelihood of fatigue is increased (Enoka & Stuart, 1992). Muscle fatigue may increase ACL loading and injury risk as it is associated with poor lower extremity biomechanics, decreased proprioception, increased joint laxity, and decreased energy absorption (Lepers, Hausswirth, Maffuletti, Brisswalter, & van Hoecke, 2000; Nyland, Shapiro, Stine, Horn, & Ireland, 1994; Rozzi, Lephart, & Fu, 1989; Skinner, 1986). Additionally, studies have shown an increase in injury rates in the later part of games, indicating that injury risk may be a function of fatigue (Gabbett, 2000; Rahnama, Reilly, & Lees, 2002). These findings suggest that if the muscle

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force necessary to perform a task can be achieved via greater strength rather than greater
activation, the likelihood of fatigue could be reduced, potentially limiting ACL injury risk.
Specifically, if the hip’s force production is derived primarily from muscle strength, less
gluteal activation would be required, potentially limiting fatigue and poor biomechanics
associated with fatigue. These studies also represent the clinical relevance in determining
the relationship between muscle strength and EMG activation amplitudes.

Comparison of the results of this study to other studies is limited, as only a few of the
previous studies have compared EMG amplitudes between individuals with different
strength values. Zeller et al. (2003) found that females demonstrated greater peak knee
valgus angles compared to males (p < 0.005) during a single-leg squat; however’ there was
not a significant difference in gluteus maximus (p = 0.199) or medius (p = 0.143) EMG
activity. Souza and Powers (2009) compared kinematics, strength, and EMG between
females with and without patellofemoral pain completing three separate tasks (running, drop
jump, and step down). Though this study did not directly compare gluteus maximus
activation and strength, the subjects with patellofemoral pain, who were statistically weaker,
also demonstrated greater gluteus maximus activation during running and step-down tasks.
However this was not the case with a drop jump task, which contradicts our findings.
Potential explanations for this difference could be that the former study’s sample was
composed exclusively of females and included an injured group. When we compared just
the stronger and weaker females in our study, there was not a significant difference in
gluteus maximus activation amplitudes regardless if they were split up based on hip
extension (p=0.125, t=-1.176) or external rotation (p=0.359, t=-0.365) strength. The findings
from this study and previous literature indicate that there could be a relationship between
muscle strength and activation patterns and that this relationship could depend on the type
of task being performed and population being tested (i.e. between genders or within
genders), which will be discussed in more detail later in this chapter.
Our study found no significant difference in EMG activation of the gluteus medius between stronger and weaker subjects. Souza and Powers (2009) didn’t note a significant difference in gluteus medius EMG amplitudes between those demonstrating patellofemoral pain and the control subjects during any task. Russell et al. (2006) also found that gluteus medius activation did not differ between genders during a single-leg drop jump task, even though females displayed greater knee valgus motion, similar to our study. Unfortunately the latter study did not investigate how strength affected kinematics or EMG. Hart et al. (2007) and Hanson et al. (2008) found significant differences in gluteus medius activity between genders. Unfortunately neither study examined knee kinematics or strength, thus it is unclear how this difference in gluteal activity influenced kinematics or was influenced by strength. The lack of comparative research indicates the need for further research in the area.

Our study also investigated the predicative capabilities of hip strength and gluteal activation in determining hip and knee kinematics. The only significant finding demonstrated that the combination of hip external rotation strength and gluteus maximus EMG amplitude predicted a limited amount of variance in peak hip external rotation ($r = 0.281$, $p = 0.039$) and knee valgus ($r = 0.348$, $p = 0.006$) angles during the loading phase of a double-leg jump landing task. However, closer analysis of the significant findings indicates that hip external rotation strength (standardized Beta-weight = 0.284) explained a greater portion of the variance in transverse plane hip motion compared to gluteal maximus EMG amplitude (standardized Beta-weight = 0.024). Similarly, hip external rotation strength (standardized Beta-weight = 0.348) was a stronger predictor of peak knee valgus angle compared to gluteal maximus EMG amplitude (standardized Beta-weight = 0.124). These findings indicate that hip external rotation strength demonstrates a limited capacity to predict hip rotation and knee valgus angles during a double-leg jump landing task. Cohen (1988), when trying to standardized the observed $r$, classifies an $r$-value of 0.10 as “small” and 0.30
as “medium.” These small standardized effect sizes and the fact that only 10% of the variance in peak knee valgus and hip internal rotation angles was explained by hip external rotation strength indicate that even though these findings are statistically significant, they may entail limited clinical or physiological significance.

Similar to the previous literature on hip strength measures and EMG, there is a limited amount of studies that were found to have investigated the predictability capabilities of hip strength on hip and knee kinematics (Claiborne, et al., 2006; Jacobs & Mattcola, 2005; Lawrence, et al., 2008; J. D. Willson, et al., 2006; J.D. Willson, et al., 2006). There are no studies to date which have investigated the capacity for gluteal EMG to predict hip and knee kinematics, so comparisons cannot be made with our study. It is important to note that all of the previous literature utilized a single-leg task in their methods, contrasting from our study. Similar to our study, Willson et al. (2006) found that greater hip external rotation strength was associated with lesser frontal plane projection angle (r=0.40) in males and females during a single leg squat. However, other studies reported that hip external rotation strength did not predict knee valgus (Claiborne, et al., 2006; Lawrence, et al., 2008). The previous literature also is contradicting with regard to the predictive capabilities of hip abduction strength and its correlation with kinematics. Similar to other studies, we found that hip abduction strength significantly did not predict knee valgus (Claiborne, et al., 2006; Willson, et al., 2006). However, Jacobs and Mattacola (2005) found a significant inverse relationship between females’ eccentric abduction peak torque and peak knee valgus angle during a single-leg hopping task (r = -0.61). The former study demonstrates the greatest correlation (r= -0.61) among all the previous literature; in fact this is the only correlation that would be classified as a large effect size according to Cohen (1988). This specific study utilized a single-leg hopping task, instead of a single-leg squat such as the other studies used or a double-leg task similar to what we used. It is plausible that this specific task was more challenging compared to the other tasks, thus possibility exploiting the weaker
subjects and creating a greater variability in kinematics between the stronger and weaker subjects.

This same theory can be applied to when discussing the lack of difference in gluteal EMG activity between those demonstrating greater and lesser strength. We theorized that activation amplitudes are related to muscle strength, and that stronger individuals would need less neural drive and activation in response to the same physical demands. However, it is likely that strength is not the only factor which influences activation, as the type of task performed introduces different physical demands. Boudreau, et al. (2009) reported that gluteus maximus and medius activity change based on the type of task being performed. Subjects used a higher percent of their peak EMG during an isometric muscle contraction when performing a single-leg squat compared a lunge and step-up-and-over task. The step-up-and-over task initiated the lowest percent of activation. This indicates that dynamic and more challenging tasks might introduce greater variability in the neural characteristics of subjects of different strength. Future research should include different tasks that might offer greater variability in not only neural activity, but kinematics as well.

All of the previous literature, in conjunction with this current study, is inconclusive when trying to determine factors that predict kinematics associated with ACL injuries. The discrepancies between these investigations are likely attributable to a number of factors including the populations being tested, the tasks being performed, and the type of strength being measured (i.e.isometric, concentric, or eccentric). Initial analyses of the results of this study indicate that there is a relationship between strength measures and kinematic values. However, with a high percentage of the strong groups being composed of males and a higher percentage of weak groups being composed of females, it is plausible that the differences in kinematics are due to sex differences and not strength differences. Secondary analyses were conducted to determine differences in strength measures and kinematics between genders and between strong and weak subjects within each gender.
Means and standard deviations for all the secondary analyses can be found in Table 5.1. Males were stronger than females for all muscle groups, even after controlling for anthropometric characteristics. Additionally, females demonstrated greater peak knee valgus and hip external rotation angles compared to males. When comparing strong vs. weak females, there was also a significant difference in all the strength measures, but the only significant kinematic finding was that females with weaker hip extensors exhibited greater peak hip external rotation. When comparing strong vs. weak males, there was also a significant difference in all the strength measures, but there was no significant difference in any kinematic measures. There were also no significant correlations between peak kinematics and muscle strength or EMG amplitudes within females or males.

Possible explanations of these secondary results include the following: 1) differences in kinematics between genders are not due to strength, but rather to inherent anatomical differences; 2) females are generally weaker than males, thus leading to gender differences in kinematics; or 3) there is not enough variability in normalized strength between the stronger and weaker individuals within a given gender to explain variance in kinematics.

It is commonly accepted that females demonstrate a different anatomical structure compared to males (Boden, et al., 2000; Hewett, et al., 2006). It is plausible that these anatomical differences contribute to the observed biomechanical differences. Females demonstrate greater anterior pelvic tilt, hip anteversion, quadriceps angle (Q-angle), tibiofemoral angle, and genu recurvatum than males (Beutler, et al., 2009; Nguyen, Boling, Levine, & Shultz, 2009; Nguyen & Shultz, 2007). Beutler, et al. (2009) found that a higher Q-angle predicted poor landing technique in males. This study didn’t report a similar influence in females; however, similar to our strength and kinematic results, it is plausible that the kinematics do not differ within genders as much as they do between genders, thus contributing to the lack of correlation in females. Nguyen et al. (2009) investigated other alignment variables that could be related to Q-angle and found that a greater tibiofemoral
angle, the natural angle in the frontal plane that is created by the anatomical axes of the tibia and femur, and femoral anteversion significantly predicted a greater Q-angle (p=0.001). This indicates that a greater Q-angle alone may not predict poor kinematics or those at a greater chance for injury, but it might be a combination of alignment characteristics. The authors conclude that all three of these variables can explain variance in the frontal plane that can predispose individuals to increased injury risks (Nguyen, et al., 2009). An anterior pelvic tilt has been found to place the hamstring in a lengthened position and shorten the rectus femoris. A lengthened hamstring may inhibit its neuromuscular facilitation by increasing the latency in the muscle and a shortened rectus femoris may allow for faster neuromuscular facilitation by decreasing the latency in the muscle (Trontelj, 1993). Both of these conditions can contribute to quadriceps dominance and affect kinematics in the sagittal plane. These studies indicate that anatomical differences can affect biomechanics and kinematics and can explain some of the variance between genders.

Males demonstrated greater strength in all measures even when body weight and segment length were taken in consideration. Strength didn’t predict or correlate to poor kinematics for males or females, but it is conceivable that if females as a collective group had comparable strength to males, their kinematics would be similar.

It is also plausible that the strength differences are responsible for the difference in kinematics between genders, but the strength variability within a given gender is not large enough to influence kinematics. However, we found that the variability among the entire sample was comparable to the variability within the male subjects in all strength measures and within the females external rotation measures. There was less variability among the females’ hip abduction and extension strength measures. It was anticipated that the standard deviations within genders were going to both be much lower than the variability among the entire sample of subjects to support the notion that limited within-gender variability can explain why there is not similar gender-specific kinematic differences among
the strong vs. weak groups. The standard deviations for each strength measure among each group can be seen in Table 5.2. These findings indicate anatomical and strength sex differences are more likely explain the results of our study and the differences in kinematics between and within genders and not a lack of variability within genders.

**Clinical Relevance**

The results of our study in combination with the previous literature are inconclusive and more research needs to be conducted to determine if the kinematic differences are due to gender differences. However, initial analysis indicates that hip strength may play a role in knee biomechanics, particularly knee valgus, a widely accepted predisposing factor to ACL injury (Claiborne, et al., 2006; Hewett, et al., 2006). Clinically, these results indicate that hip strengthening may be an important component to include in preventative ACL protocols and rehabilitation programs, particularly in females who are lacking comparable hip strength to their male counterparts. This information can also potentially be generalized to general knee injury prevention programs and correction of poor knee biomechanics analyzed during functional movements.

The results also indicate that hip strength may play a role in gluteal EMG amplitudes during functional movement, in that weaker individuals demonstrate greater EMG amplitudes compared to stronger individuals. Previous literature has indicated that an increase in neural drive can increase fatigue and in turn, fatigue has been associated with increased injury rates (Enoka & Stuart, 1992; Gabbett, 2000; Rahnama, Reilly, & Lees, 2002). Hypothetically if individuals can control their lower extremity primarily with their strength instead of neural drive, fatigue could be reduced, thus potentially reducing the risk of injury. More research needs to conducted in order to support this hypothesis, but these findings indicate the clinical importance strong hip musculature. As future research
continues to evaluate and determine predisposing factors that influence biomechanics associated with ACL tears, better ACL prevention programs can be created.

**Limitations**

This study was not without limitations. Our study focused on healthy active individuals who were between the ages of 18-30, thus generalization of our results is limited to this population. However, we believe these findings have clinical relevance as ACL injuries commonly occur in this population. It is unclear if these findings can be generalized to those individuals who are at the greatest risk of ACL injury: females and athletes participating in activities that have a high incidence of cutting, sudden deceleration (Griffin, et al., 2000; Griffin, et al., 2006). Additionally the kinematic data was obtained during a double-leg jump landing task, and it is unclear if this task is a true representation of a scenario in which ACL injury occurs. A cutting task or a dynamic single-leg task might have been a better representation.

Another possible limitation is that the same EMG site was used to measure gluteus maximus activation when conducting the MVIC trials for hip extension and external rotation. We are unaware of any previous literature regarding different EMG electrode sites for assessing the various functions of the gluteus maximus (i.e. different areas of the gluteus maximus), but during data collection a definitive increase in EMG amplitude could be seen on the oscilloscope during MVICs for hip extension, but this was not always true for hip external rotation. It is uncertain if a different EMG site would have changed the results for this study.

**Conclusions**

The only definitive conclusions that can be made are that males have stronger hip extensors, external rotators, and abductors, and land from a double-leg jump landing task with lesser peak knee valgus and hip external rotation compared to females. The only
kinematic difference that still exists when comparing within genders is that females with weaker hip extensors, land with greater hip external rotation. This study indicates that there are kinematic and neuromuscular differences between males and females, but these differences are not always prevalent within genders. Females presented with greater knee valgus, a widely accepted predisposing factor to ACL injury in the current literature (Claiborne, et al., 2006; Hewett, et al., 2006). This finding offers support to reasoning why females injure their ACL a higher rate, but research needs to continue to be done investigating why some females and males tear their ACLs and others do not. Specifically research needs to be conducted determining if there is a true relationship between strength and neural activity; if different static and dynamic tasks change the predictably capabilities of hip strength and peak activation; if the difference in kinematics between genders is due to sex differences; if displacements have a role in predicting kinematics; and if females’ biomechanics are altered such that they are similar to males if their gluteal strength is increased through training in a prospective study.
### Table 5.1. Secondary Analyses

<table>
<thead>
<tr>
<th></th>
<th>Differences between genders</th>
<th>Differences between females</th>
<th>Differences between males</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Hip Extension</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Strength (%BW*Ht)</td>
<td>0.111 ± 0.024*</td>
<td>0.091 ± 0.017</td>
<td>0.110 ± 0.008*</td>
</tr>
<tr>
<td>Peak Hip Internal Rotation(°)†</td>
<td>-2.60 ± 9.72*</td>
<td>-7.60 ± 8.51</td>
<td>-3.24 ± 7.62*</td>
</tr>
<tr>
<td>Gluteus Maximus Peak EMG</td>
<td>128.74 ± 275.15</td>
<td>200.71 ± 447.60</td>
<td>112.95 ± 135.19</td>
</tr>
<tr>
<td>Amplitude (%MVIC)</td>
<td>79.32*</td>
<td>467.07</td>
<td>199.99</td>
</tr>
<tr>
<td><strong>Hip External Rotation</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Strength (%BW*Ht)</td>
<td>0.046 ± 0.009*</td>
<td>0.0410 ± 0.007</td>
<td>0.049 ± 0.004*</td>
</tr>
<tr>
<td>Peak Hip Internal Rotation(°)†</td>
<td>-2.60 ± 9.72*</td>
<td>-7.60 ± 8.51</td>
<td>-4.38 ± 8.07</td>
</tr>
<tr>
<td>Gluteus Maximus Peak EMG</td>
<td>128.74 ± 275.15</td>
<td>216.65 ± 273.16</td>
<td>130.11 ± 146.11</td>
</tr>
<tr>
<td>Amplitude (%MVIC)</td>
<td>79.32*</td>
<td>467.07</td>
<td>197.33</td>
</tr>
<tr>
<td><strong>Hip Abduction</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Strength (%BW*Ht)</td>
<td>0.113 ± 0.030*</td>
<td>0.089± 0.017</td>
<td>0.109 ± 0.008*</td>
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<tr>
<td>Peak Hip Adduction (°)#</td>
<td>2.40 ± 7.86</td>
<td>3.71 ± 7.00</td>
<td>5.16 ± 6.11</td>
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<tr>
<td>Gluteus Medius Peak EMG</td>
<td>98.68 ± 149.44</td>
<td>142.20 ± 193.14</td>
<td>104.58 ± 101.33</td>
</tr>
<tr>
<td>Amplitude (%MVIC)</td>
<td>103.14*</td>
<td>155.34</td>
<td>106.79</td>
</tr>
</tbody>
</table>

† Negative value indicates hip external rotation and a positive value indicates hip internal rotation
‡ Negative value indicates knee valgus and a positive value indicates knee varus
# Negative value indicates hip abduction and a positive value indicate hip adduction
*p<0.05
Table 5.2. Standard Deviations within females, males, and the entire sample strength measures

<table>
<thead>
<tr>
<th></th>
<th>Within Males</th>
<th>Within Females</th>
<th>Entire Sample</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip Extension</td>
<td>0.024</td>
<td>0.016</td>
<td>0.023</td>
</tr>
<tr>
<td>Hip Abduction</td>
<td>0.030</td>
<td>0.017</td>
<td>0.027</td>
</tr>
<tr>
<td>Hip External Rotation</td>
<td>0.009</td>
<td>0.007</td>
<td>0.009</td>
</tr>
</tbody>
</table>
Appendix: Manuscript

The influence of hip strength on gluteal muscle activation and lower extremity kinematics

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The influence of hip strength on gluteal muscle activation and lower extremity kinematics

Abstract

Objective: To evaluate the influence of hip strength on kinematic ACL injury risk factors and gluteal activation. Design: Cross-sectional. Setting: Research laboratory. Participants: Eighty-two healthy volunteers. Outcome Measures: Hip extension, external rotation, and abduction strength; gluteus maximus and medius electromyography (EMG); knee valgus, hip adduction, and hip internal rotation angles. Results: Peak knee valgus (p=0.016) and hip external rotation (p=0.023) angles were greater in individuals with weaker hip external rotators. Gluteus maximus EMG was greater in individuals with weaker hip extensors (p=0.031) and external rotators (p=0.043). Hip external rotation strength and gluteus maximus EMG amplitude predicted 7.9% and 12.1% of the variance in peak hip rotation (R=0.281, p=0.039.) and knee valgus angles (R=0.348, p=0.006). Conclusion: Hip external rotation strength influences hip and knee kinematics related to ACL injury, thus increasing gluteal strength may be an important addition to ACL injury prevention programs. Key Words: ACL, strength, EMG, knee valgus
Introduction

Annually Americans spend $1.7 billion repairing some of the estimated 250,000 anterior cruciate ligament (ACL) tears that result in multiple short and long term effects including financial, psychological, and physical stresses. Therefore, it is important to understand modifiable factors that predispose individuals to ACL injury so that effective injury prevention protocols can be developed. One biomechanical risk factor that has been widely researched is the angle of knee valgus that occurs during functional movement. Knee valgus moment combined with an anterior shear force creates a larger strain on the cadaveric ACL compared to isolated anterior shear force. The load in the knee can be increased 6x by increasing knee valgus just 5° from neutral alignment. Hewett et al. demonstrated that greater external knee valgus angles, moments, and side-to-side differences in females had a predictive value of 0.88 for ACL tears. Knee valgus motion results from a combination of femoral adduction and internal rotation, along with tibial external rotation and abduction. The gluteal muscles eccentrically resist hip adduction and internal rotation, thus a weakened gluteus medius may permit greater hip adduction and a weakened gluteus maximus may permit greater hip internal rotation, resulting in greater knee valgus during dynamic tasks.

The literature regarding the influence of gluteal muscle strength on frontal plane knee motion is inconsistent. Some studies have reported that greater knee valgus is associated with muscle weakness, while other studies have not found a significant association in static or dynamic assessments. However, dynamic tasks such as landing do not typically require maximal muscular force production, thus peak strength measures may not adequately predict lower extremity kinematics during these tasks. Bell et al. concluded that hip muscle strength does not influence lower extremity kinematics and instead suggested that muscle activation amplitudes and patterns could explain the differences in lower extremity kinematics. Therefore, possessing a high level of gluteal muscle strength does not necessitate a high level of gluteal activity during a dynamic task. It is plausible that muscle activation (i.e. EMG activity)
may be a more influential determinant of lower extremity kinematics than muscle strength, and may partially explain the discrepancies in the literature regarding the influence of gluteal strength on knee valgus motion. Similar to gluteal strength, however, the literature regarding the influence of gluteal activation on knee valgus is inconclusive, as some studies demonstrate a difference in gluteal activation between those demonstrating different kinematics, while others do not.

While numerous studies have reported the influence of gluteal muscle force or activity in isolation, we are unaware of any previous literature which has evaluated these factors concomitantly. The conflicting evidence in this area suggests that a combination of these two factors may be more effective for elucidating the role of the gluteal musculature in controlling lower extremity kinematics and ACL injury risk. The primary purpose of this study was to investigate the influence of gluteal muscle strength on gluteal EMG amplitudes, peak knee valgus, peak hip adduction, and peak hip internal rotation during a double-leg jump landing task. A secondary purpose was to determine if the combination of gluteal strength and EMG activity predicts knee and hip landing kinematics more effectively than either variable in isolation. We hypothesized that there would be no difference in peak knee valgus, hip adduction, and hip internal rotation between those demonstrating high versus low isometric hip abduction, external rotation, and extension strength, and that weaker individuals would display greater gluteal EMG activity during landing. We also hypothesized that the linear combination of gluteal strength and activation would predict peak hip internal rotation, hip adduction, and knee valgus angles.

Methodology

This cross sectional investigation utilized a casual-comparative design to evaluate the influence of gluteal strength and EMG amplitudes on peak knee valgus, hip adduction, and internal rotation angles. Isometric hip extension, abduction, and external rotation strength were measured first via a hand-held dynamometer (Chatillon CSD 300, Amteck, Inc., Largo, FL) in a
randomized order. Lower extremity kinematics and gluteal EMG were then measured as the subjects performed a double-leg drop jump landing task. All data were sampled from the dominant limb, defined as the leg used to kick a ball for maximal distance.

Participants

A convenience sample of 82 healthy, physically active volunteers (41 males: age = 21.2 ± 2.3, height = 179.0 ± 7.0 cm, mass = 78.5 ± 13.9 kg and 41 females: age = 21.4 ± 3.2, height = 166.4 ± 6.5 cm, mass = 62.5 ± 9.8 kg) participated in this study. Subjects were required to have no history of lower extremity surgery, ACL injury, chronic or neurological disorders, or lower extremity injury in the 6 months prior to data collection that prevented them from carrying out activities of everyday life, and to be physically active, participating in at least 30 minutes of physical activity 3 times per week. All subjects read and sign an informed consent document approved by the university's institutional review board.

Procedures

EMG electrodes (DelSys Bagnoli-8, DelSys Inc., Boston, MA: interelectorde distance = 10 mm; amplification factor = 10,000; CMMR at 60 Hz > 80 dB; input impedance > 10^15 ohms/0.2 pF) were placed over the gluteus maximus and gluteus medius muscle bellies parallel with the muscle fibers. Electrode placement sites were prepared by shaving, abrading, and cleaning the skin with alcohol pads to reduce impedance, and were confirmed via manual muscle tests. Electrode placement on the gluteus maximus was located at the proximal 1/3 of the distance between the greater trochanter of the femur and the spinous process of the S2 vertebrae. The electrodes placed on the gluteus medius were positioned so that they were located at the distal 1/3 of the distance between the iliac crest and greater trochanter of the femur. A reference electrode was placed over the proximal antero-medial tibia.
Maximal voluntary isometric contractions (MVICs) for the dominant limb hip abductors (ICC(2,1) = 0.97), extensors (ICC(2,1) = 0.95), and external rotators (ICC(2,1) = 0.96) were assessed via a hand-held dynamometer (Chatillon CSD 300, Amteck, Inc., Largo, FL) by the same investigator (MFN) to enhance reliability. Three 5s trials were performed with 1 minute rest between trials to minimize the risk of fatigue. Subjects were positioned as recommended by Hislop, Mongomery, Connelly, & Daniels\textsuperscript{17} and received verbal encouragement as they produced maximal force against the dynamometer. Peak force (N) obtained from the dynamometer was multiplied by segment length (m) to calculate peak torque (N·m) which was normalized to the product of weight and height.

Subjects performed a double-leg jump landing from a box 30 cm in height positioned at a distance equal to 50% of their height away from a non-conductive force plate (Model 4060-NC; Bertec Corporation, Columbus, OH). Following 3 practice trials, subjects performed 5 successful trials with 30 seconds between each trial to minimize the risk of fatigue. Successful trials required the subjects to land with the dominant foot completely on the force plate and the non-dominant foot completely off the force plate and to immediately jump vertically for maximum height following initial ground contact. Unsuccessful trials were repeated until 5 successful trials were obtained.

An electromagnetic motion capture system (Motion Star, Ascension Technology Corp., Burlington, VT) was used to assess knee and hip kinematics during the jump landing task. Three-dimensional coordinate data were sampled at 120 Hz while force plate and EMG data were sampled simultaneously at 1,200 Hz. Global and segment axis systems were established using a right-hand coordinate system where X was positive anteriorly/forward, Y was positive medially/leftward, and Z was positive superiorly/upward. Electromagnetic sensors were placed on the pelvis, thigh, and shank with double-sided tape and secured with pre-wrap and tape. A segment-linkage model of the dominant lower extremity was generated by digitizing anatomical landmarks. The location of the hip joint center was estimated as a function of the distance
between the digitized anterior superior iliac spines. The knee and ankle joint centers were defined as the midpoint between the digitized lateral and medial femoral condyles and lateral and medial malleoli, respectively.

Data Reduction and Analysis

Knee and hip angular kinematics were calculated using Euler angles rotated in a Y X' Z'' sequence. Motion about the knee was defined as the shank reference frame relative to the thigh reference frame such that flexion, varus, and internal rotation represented positive values. Motion about the hip was defined as the thigh reference frame relative to the pelvis reference frame such that extension, adduction, and internal rotation represented positive values. Three-dimensional coordinate data were lowpass filtered at 12 Hz (4th order zero-phase-lag Butterworth) and were time-synchronized to the EMG and kinetic data and re-sampled to 1,200 Hz via linear interpolation. EMG data were corrected for DC bias and band-pass (20-350 Hz) and notch (59.5-60.5 Hz) filtered (4th order zero-phase-lag Butterworth), and smoothed using a 25ms sliding window function via custom software (LabVIEW, Natonal Instruments, San Antonio, TX).

Peak hip adduction, hip internal rotation, and knee valgus angles, along with the mean EMG amplitudes of the gluteus medius and maximus were calculated during the loading phase of the landing task. The loading phase was defined as the time interval from initial ground contact to maximum knee flexion. Initial ground contact was defined as the instant at which the vertical ground reaction force exceeded 10N. The peak EMG amplitude was determined over the middle 3 seconds of the MVIC trials, and was averaged over the 3 MVIC trials per muscle group. These values were used to normalize EMG amplitudes of the respective muscles during the jump landing task. The EMGs during the jump landing task were normalized to the MVICs and reported as %MVIC. During data collection when conducting the MVIC trials for hip extension and external rotation there was not always a definitive spike in EMG amplitude during
the external rotation trials unlike the hip extension and abduction trials. Because of this observation, we normalized all gluteus maximus EMG amplitudes to the extension MVIC EMG amplitudes. Strength data were arranged into tertiles (n = 27 for highest and lowest tertiles and 28 for middle), thus creating groups with high vs. low strength of the respective muscle groups, and independent-samples t-tests were used to determine if the strength differed between the highest and lowest tertiles. Gluteus medius and maximus EMG amplitudes and peak knee valgus, hip adduction, and hip internal rotation angles were compared between the high and low strength groups via independent-samples t-tests. Multiple linear regression analyses were used to evaluate the relationships between peak kinematic values during the jump landing task and the linear combinations of muscle strength and EMG amplitude. For all linear regression analyses the muscle strength data was the first predictor variable entered, followed by the EMG data. All data were analyzed using PASW 18.0 (Chicago, IL) statistical software with statistical significance established a priori as α ≤ 0.05.

Results

The high and low strength tertiles were significantly different for hip external rotation (p < 0.001), hip abduction (p < 0.001), and hip extension (p < 0.001), verifying the presence of “strong” and “weak” groups. Means and standard deviations for each strength comparison, and demographics for each tertile are detailed in Table 1.

Hip Strength and Kinematics

There were no significant differences in peak hip adduction, hip internal rotation, or knee valgus angles between those demonstrating high vs. low isometric hip abduction or extension strength. However, individuals with greater hip external rotation strength demonstrated lesser
peak hip external rotation ($p = 0.0236$) and knee valgus ($p = 0.0158$) compared to those with weaker hip external rotators (Table 1).

*Hip Strength and EMG Amplitude*

Gluteus maximus EMG amplitudes were significantly smaller in individuals with stronger hip extensors ($p = 0.031$) and external rotators ($p = 0.043$). No significant difference was found in EMG amplitudes between the strong and weak hip abduction groups (Table 1).

*Regressions*

The linear a combination of hip external rotation strength and peak gluteus maximus EMG amplitude predicted 7.9% of the variance in peak hip rotation angle ($r = 0.281$, $p = 0.039$) and 12.1% of the variance in peak knee valgus angle ($r = 0.348$, $p = 0.006$). Analysis of the regression models indicates that hip external rotation strength (standardized Beta-weight = 0.284) explained a greater portion of the variance in peak hip external rotation angle compared to gluteal maximus activation (standardized Beta-weight = 0.024). Similarly, hip external rotation strength (standardized Beta-weight = 0.348) was a stronger predictor of peak knee valgus angle compared to gluteal maximus activation (standardized Beta-weight = 0.124). Descriptions for each regression model are presented in Table 2.

*Discussion*

Our investigation compared peak hip and knee kinematics and EMG activation of the gluteal muscles between subjects demonstrating greater and lesser isometric hip extension, external rotation, and abduction strength. We found that subjects with greater hip external rotation strength demonstrated lesser peak hip external rotation and knee valgus angles; those subjects demonstrating greater hip extension strength demonstrated lesser gluteus maximus EMG amplitudes during landing. Additionally, the combination of hip external rotation strength
and gluteus maximus EMG amplitude explained a significant, though limited, amount of variance in peak hip rotation and knee valgus angles.

Individuals with greater isometric hip external rotation strength demonstrated lesser peak knee valgus angles compared to weaker individuals. This difference suggests that subjects with stronger hip external rotators possess a better capacity for resisting knee movement in the frontal plane, which agrees with some of the previous literature. Conversely, Lawrence et al. found no significant difference in peak knee valgus angles between females demonstrating greater vs. lesser hip external rotation strength. However, when we limited our statistical analyses to stronger vs. weaker females, similar to Lawrence’s study, there was not a significant difference in peak knee valgus angles (p = 0.387). This finding agrees with the former study and indicates that the difference in kinematics maybe due to sex differences, not necessarily strength differences.

Individuals with stronger hip external rotators also displayed lesser peak hip external rotation angles compared to weaker individuals. Hip internal rotation is one component of knee valgus Therefore, greater hip internal rotation would seemingly be associated with greater knee valgus. As such, we hypothesized that individuals with greater hip external rotation strength would demonstrate less hip internal rotation and, therefore, less knee valgus. However, our data indicates that individuals with greater hip external rotation strength demonstrated less knee valgus, but also demonstrated less peak external rotation (i.e. a more internally rotated hip) compared to those with weaker external rotators. In an attempt to explain these seemingly contradictory results, we compared hip internal rotation displacement between the stronger and weaker external rotation groups and found that the stronger group demonstrated less displacement (6.79° ± 1.16) compared to the weaker group (10.18° ± 1.02) (p = 0.017). This indicates that the weaker group started in a more externally rotated position but moved through a greater range of motion toward hip internal rotation, likely explaining the greater peak knee valgus angle in weaker individuals.
The previous results suggest that gluteal strength may influence lower extremity
kinematics. However, Bell et al.\textsuperscript{10} found that individuals with excessive medial knee
displacement had significantly greater hip extension and hip external rotation strength compared
to those subjects displaying a neutral frontal plane angle, and speculated that possessing
adequate strength to control the lower extremity during dynamic tasks may not necessitate the
appropriate level of muscle activation. Dynamic tasks such as landing do not require maximal
muscular force production, thus peak strength measures may not adequately predict kinematics.
Instead, the level of muscle activation may be a more important determinant of lower extremity
motion.\textsuperscript{10} Our study found that individuals with greater hip extension and external rotation
strength demonstrated lesser gluteus maximus EMG amplitudes compared to weaker
individuals. Therefore, a weaker muscle requires greater activation to control the lower
extremity compared to a stronger muscle. This greater neural drive requires a greater amount
of energy substrate, thus increasing the likelihood of fatigue.\textsuperscript{19} Muscle fatigue may increase
ACL loading and injury risk as it is associated with poor lower extremity biomechanics,
decreased proprioception, increased joint laxity, and decreased energy absorption.\textsuperscript{20-24} An
increase in injury rates in the later part of games indicates that injury risk may be a function of
fatigue.\textsuperscript{25} These findings suggest that if force production can primarily be achieved through
greater strength and the less neural drive, the risks of fatigue and ACL injury may be reduced.
Specifically, if the hip’s force production primarily came from muscle strength, less gluteal
activation would be required, potentially limiting fatigue and poor biomechanics associated with
fatigue.

Comparison of our results to other studies is limited, as only a few previous studies have
investigated differences in EMG amplitudes among those with different strength values. Souza
and Powers\textsuperscript{26} compared kinematics, strength, and EMG amplitudes between females with and
without patellofemoral pain during three tasks (running, drop jump, and step down). Though
this study did not directly compare gluteus maximus activation and strength, the subjects with
patellofemoral pain, who were statistically weaker, also demonstrated greater gluteus maximus activation during running and step-down tasks. However this was not the case with a drop jump task, which contradicts our findings. When we limited our comparisons to stronger vs. weaker females, similar to the former study, there was not a significant difference in gluteus maximus activation amplitudes (p = 0.244). Our study also found no significant difference in EMG activation of the gluteus medius between the stronger and weaker subjects. To our knowledge there has not been another study that primarily investigated the relationship between hip strength and gluteus medius activity. There have been other studies that investigated the relationship between EMG amplitudes and kinematics, but the results are inconclusive.\textsuperscript{18-20} The findings from this study and previous literature indicate that there could be a relationship between muscle strength and activation, but that this relationship could depend on the type of task being performed and population being tested (i.e. between genders or within genders).

Our study also investigated the predicative capabilities of hip strength and gluteal activation in determining hip and knee kinematics. The only significant finding demonstrated that hip external rotation strength predicted a limited amount of variance in peak hip external rotation and knee valgus angles. Similar to the previous literature on hip strength measures and EMG activation, there is a limited amount of studies which investigated the of the influence of hip strength on hip and knee kinematics.\textsuperscript{8,9,12,13,27} There are no studies to date known to have investigated the of the relationship between gluteal EMG and kinematics, so comparisons cannot be made with our study. It is important to note that all of the previous literature utilized single-leg tasks as opposed to the double-leg task used in our study. Similar to our study, Willson et al.\textsuperscript{9} found that greater hip external rotation strength was associated with lesser frontal plane knee angle (r = 0.40) in males and females during a single-leg squat. However, other studies reported that hip external rotation strength did not predict knee valgus.\textsuperscript{8,13} The previous literature also is contradicting regarding the relationship between hip abduction strength and lower extremity kinematics. Similar to other studies, we found that hip abduction strength did not
predict knee valgus motion. However, Jacobs and Mattacola found a significant inverse relationship between females’ eccentric abduction peak torque and peak knee valgus angle during a single-leg hopping task \( (r = -0.61) \). This study utilized a single-leg hopping task instead of a single-leg squat such as the other studies used or a double-leg task similar to what we used. It is plausible that this specific task was more challenging compared to the other tasks, thus possibility exploiting the weaker subjects and creating a greater variability in kinematics between the stronger and weaker subjects.

We hypothesized that EMG amplitudes would be related to muscle strength, with stronger individuals requiring less neural drive to perform the same task compared to weaker individuals. However, both strength and the type of task being performed influence EMG amplitudes. Boudreau, et al. reported that gluteus maximus and medius activity change based on the type of task being performed. Subjects used a higher percent of their peak EMG during an isometric muscle contraction during a single-leg squat compared a lunge and step-up-and-over task. The step-up-and-over task initiated the lowest percent of activation. This indicates that tasks which are more dynamic and challenging compared to double-leg landings might more clearly elucidate the influence of gluteal strength on EMG amplitudes.

Initial analyses of the results of this study indicate that there is a relationship between strength measures and kinematic values. However closer inspection of our sample indicates that the strong groups were primarily composed of males while the weak groups were primarily composed of females, even after controlling for anthropometric differences (Table 1). It is plausible that the differences in kinematics, particularly knee valgus angle, are due to sex differences and not strength differences. Secondary analyses were conducted to compare strength and kinematics between and within genders. Males were significantly stronger for each muscle group (Table 1). Accordingly, females demonstrated greater peak knee valgus and hip external rotation angles. When comparing the strong and weak females, there was also a significant difference in all the strength measures, but the only significant kinematic finding was
that females demonstrating weaker hip extension strength exhibited greater peak hip external rotation. When comparing the strong and weak males, there was also a significant difference in all the strength measures, but there was no significant difference in any kinematic measures. There also were also no significant correlations between peak kinematics and muscle strength or EMG amplitudes in females or males, respectively.

Possible explanations of these secondary results include the following: 1) differences in kinematics between genders are not due to strength, but rather to inherent anatomical differences; 2) there is not enough variability in normalized strength between the stronger and weaker individuals within a given gender to explain variance in kinematics; or 3) females are generally weaker than males, thus leading to gender differences in kinematics.

It is commonly accepted that females demonstrate a different anatomical structure compared to males. It is plausible that these anatomical differences contribute to the observed biomechanical differences. Females demonstrate greater anterior pelvic tilt, hip anteversion, quadriceps angle (Q-angle), tibiofemoral angle, and genu recurvatum than males. Beutler, et al. found that a larger Q-angle predicted poor landing technique in males. This study didn’t report for a similar influence in females; however, similar to our strength and kinematic results, it is plausible that the kinematic do not differ within genders as much as they do between genders, thus contributing to the lack of correlation in females.

Nguyen et al. investigated other alignment variables that could be related to Q-angle and found that a greater tibiofemoral angle, the natural angle in the frontal plane that is created by the anatomical axes of the tibia and femur, and femoral anteversion significantly predicted a greater Q-angle (p=0.001). This indicates that a greater Q-angle alone may not predict poor kinematics or those at a greater chance for injury, but it might be a combination of alignment characteristics. The authors conclude that all three of these variables can explain variance in the frontal plane that can predispose individuals to increased injury risks. An anterior pelvic tilt has been found to place the hamstring in a lengthened position and shorten the rectus femoris.
A lengthened hamstring may inhibit its neuromuscular facilitation by increasing the latency in the muscle and a shortened rectus femoris may allow for faster neuromuscular facilitation by decreasing the latency in the muscle. Both of these conditions can contribute to quad dominance and affect kinematics in the sagittal plane. These studies indicate that anatomical differences can affect biomechanics and kinematics and can explain some of the variance between genders.

Males demonstrated greater strength in all measures even when body weight and segment length were taken in consideration. Strength didn’t predict or correlate to poor kinematics for males or females, but it is conceivable that if females as a collective group had comparable strength to males, their kinematics would be similar.

It is also plausible that the strength differences are responsible for the difference in kinematics between genders, but the strength variability within a given gender is not large enough to influence kinematics. We found that the variability among the entire sample was comparable to the variability within the male subjects in all strength measures and within the females' hip abduction and extension strength measures. There was less variability among the females’ hip abduction and extension strength measures. It was anticipated that the standard deviations within genders were going to both be much lower than the variability among the all the subjects to support the notion that limited within-gender variability is the reason there is not similar gender-specific kinematic differences among the strong vs. weak groups. These findings indicate anatomical and strength sex differences more likely explain the results of our study and the differences in kinematics between and within genders and not a lack of variability within genders.

Limitations

This study was not without limitations. Our study focused on healthy active individuals who were between the ages of 18-30, thus generalization of our results is limited to this
population. However, we believe these findings have clinical relevance as ACL injuries commonly occur in this population. It is unclear if these findings can be generalized to those individuals who are at the greatest risk of ACL injury: females and athletes participating in activities that have a high incidence of cutting, sudden deceleration and pivoting.\textsuperscript{34,35}

Additionally, the kinematic data were obtained during a double-leg jump landing task, and it is unclear if this task is a true representation of a scenario in which ACL injury occurs. A cutting task or a dynamic single-leg task might have been a better representation.

Another possible limitation is that the same EMG site was used to measure gluteus maximus activation when conducting the MVIC trials for hip extension and external rotation. We are unaware of any previous literature regarding different EMG electrode sites for assessing the various functions of the gluteus maximus (i.e. different areas of the gluteus maximus), but during data collection a definitive increase in EMG amplitude could be seen on the oscilloscope during MVICs for hip extension, but this was not always true for hip external rotation. It is uncertain if a different EMG site would have changed the results for this study.

Conclusions

The only definitive conclusions that can be made are that males have stronger hip extensors, external rotators, and abductors, and land from a double-leg jump landing task with lesser peak knee valgus and hip external rotation compared to females. The only kinematic difference that still exists when comparing within genders is that females with weaker hip extensors, land with greater hip external rotation. This study indicates that there are kinematic and neuromuscular differences between males and females, but these differences are not always prevalent within genders. Females presented with greater knee valgus, a widely accepted predisposing factor to ACL injury in the current literature.\textsuperscript{2,8} This finding offers support to reasoning why females injure their ACL a higher rate, but research needs to continue to be done investigating why some females and males tear their ACLs and others do not. Specifically
research needs to be conducted determining if there is a true relationship between strength and neural activity; if different static and dynamic tasks change the predictably capabilities of hip strength and peak activation; if the difference in kinematics between genders is due to sex differences; if displacements have a role in predicting kinematics; and through a prospective study if females’ gluteal strength is increased through training if their biomechanics are altered such that they are similar to males prior to training.
Table 1. Means and standard deviations for the differences in strength, EMG activity, and kinematics between tertiles and within genders.

<table>
<thead>
<tr>
<th></th>
<th>Differences between tertiles</th>
<th>Differences between females</th>
<th>Differences between males</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>High Tertile (27)</td>
<td>Low Tertile (27)</td>
<td>Observed Power</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>Stronger (14)</td>
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<td>Weaker (14)</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>Stronger (14)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Weaker (14)</td>
</tr>
<tr>
<td><strong>Hip Extension</strong></td>
<td>( 19 males, 8 females)</td>
<td>(7 males, 20 females)</td>
<td></td>
</tr>
<tr>
<td>Strength (%BW*Ht)</td>
<td>0.0575 ± 0.0057*</td>
<td>0.0363 ± 0.0030</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>0.110 ± 0.008*</td>
</tr>
<tr>
<td>Valgus (°)†</td>
<td>-15.46 ± 9.00</td>
<td>-19.27 ± 10.61</td>
<td>0.388</td>
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<tr>
<td></td>
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<td></td>
<td>-17.68 ± 6.10</td>
</tr>
<tr>
<td>Rotation (°)†</td>
<td>-3.77 ± 8.95</td>
<td>-7.61 ± 9.04</td>
<td>0.427</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>-3.24 ± 7.62*</td>
</tr>
<tr>
<td>Gluteus Maximus Peak</td>
<td>117.72 ± 50.08*</td>
<td>322.22 ± 556.49</td>
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</tr>
<tr>
<td>EMG Amplitude (%MVIC)</td>
<td></td>
<td></td>
<td>200.71 ± 199.99</td>
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<td>447.60 ± 759.48</td>
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<td>112.95 ± 53.63</td>
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<tr>
<td><strong>Hip External Rotation</strong></td>
<td>(22 males, 5 females)</td>
<td>(4 males, 23 females)</td>
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<tr>
<td>Strength (%BW*Ht)</td>
<td>0.1272±0.0170*</td>
<td>0.0785 ± 0.0092</td>
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<tr>
<td></td>
<td></td>
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<td>0.049 ± 0.004*</td>
</tr>
<tr>
<td>Valgus (°)†</td>
<td>-14.65 ± 11.28*</td>
<td>-21.17 ± 10.39</td>
<td>--</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>-18.53 ± 8.44</td>
</tr>
<tr>
<td>Rotation (°)†</td>
<td>-3.21 ± 9.92*</td>
<td>-8.39 ± 8.75</td>
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<td>-4.38 ± 8.07</td>
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<tr>
<td>Gluteus Maximus Peak</td>
<td>117.39 ± 63.20*</td>
<td>307.00 ± 559.75</td>
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<tr>
<td>EMG Amplitude (%MVIC)</td>
<td></td>
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<td>216.65 ± 197.33</td>
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<td></td>
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<td>273.16 ± 545.06</td>
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<td>130.11 ± 71.20</td>
</tr>
<tr>
<td><strong>Hip Abduction</strong></td>
<td>(21 males, 6 females)</td>
<td>(9 males, 18 females)</td>
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</tr>
<tr>
<td>Strength (%BW*Ht)</td>
<td>0.1317 ± 0.0235*</td>
<td>0.0754 ± 0.0065</td>
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<tr>
<td></td>
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<td>0.109 ± 0.008*</td>
</tr>
<tr>
<td>Valgus (°)†</td>
<td>-15.34 ± 9.00</td>
<td>-18.92 ± 9.34</td>
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<td>-21.04 ± 8.83</td>
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<tr>
<td>Frontal Plane (°)†</td>
<td>1.98 ± 8.31*</td>
<td>3.19 ± 6.77</td>
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<tr>
<td></td>
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<td>5.16 ± 6.11</td>
</tr>
<tr>
<td>Gluteus Medius Peak</td>
<td>101.08 ± 117.15</td>
<td>155.10 ± 177.36</td>
<td>0.367</td>
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<tr>
<td>EMG Amplitude (%MVIC)</td>
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<td>142.20 ± 106.79</td>
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<td></td>
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<td></td>
<td>193.14 ± 234.90</td>
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<tr>
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<td>104.58 ± 157.51</td>
</tr>
</tbody>
</table>

† Negative value indicates hip external rotation and a positive value indicates hip internal rotation
‡ Negative value indicates knee valgus and a positive value indicates knee varus
# Negative value indicates hip abduction and a positive value indicate hip adduction
* p<0.05
Table 2. Predictability of hip and knee kinematics based on hip strength and gluteal activation

<table>
<thead>
<tr>
<th>Variable</th>
<th>Parameter</th>
<th>Standard Error</th>
<th>Standardized Coefficient</th>
<th>t Value</th>
<th>p Value for individual predictors</th>
<th>R²</th>
<th>Regression p Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Hip Rotation</strong></td>
<td></td>
<td></td>
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<tr>
<td>Intercept</td>
<td>-18.248</td>
<td>5.262</td>
<td>-0.348</td>
<td>-3.468</td>
<td>0.001</td>
<td>0.079</td>
<td>0.039</td>
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<td>Hip external rotation strength</td>
<td>280.880</td>
<td>108.367</td>
<td>0.284</td>
<td>2.592</td>
<td>0.011</td>
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<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.001</td>
<td>0.003</td>
<td>0.024</td>
<td>0.222</td>
<td>0.825</td>
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<tr>
<td>Intercept</td>
<td>-13.626</td>
<td>5.041</td>
<td>-0.270</td>
<td>-2.703</td>
<td>0.008</td>
<td>0.039</td>
<td>0.209</td>
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<tr>
<td>Hip extension strength</td>
<td>82.725</td>
<td>46.705</td>
<td>0.202</td>
<td>1.771</td>
<td>0.080</td>
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<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.069</td>
<td>0.315</td>
<td>0.025</td>
<td>0.219</td>
<td>0.827</td>
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<tr>
<td><strong>Knee Valgus</strong></td>
<td></td>
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<tr>
<td>Intercept</td>
<td>-35.703</td>
<td>5.750</td>
<td>-0.621</td>
<td>-6.210</td>
<td>&lt; 0.001</td>
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<td>Hip external rotation strength</td>
<td>383.977</td>
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<td>3.243</td>
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<tr>
<td>Peak Gluteus Maximus EMG (%MVIC)</td>
<td>0.044</td>
<td>0.003</td>
<td>0.124</td>
<td>1.156</td>
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<tr>
<td>Intercept</td>
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<td>-0.534</td>
<td>-5.341</td>
<td>&lt; 0.001</td>
<td>0.036</td>
<td>0.237</td>
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<td>Hip abduction strength</td>
<td>72.899</td>
<td>42.689</td>
<td>0.189</td>
<td>1.708</td>
<td>0.092</td>
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<tr>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td>0.140</td>
<td>0.872</td>
<td>0.018</td>
<td>0.161</td>
<td>0.873</td>
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<tr>
<td>Intercept</td>
<td>-28.871</td>
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<td>-0.516</td>
<td>-5.167</td>
<td>&lt; 0.001</td>
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<td>Hip extension strength</td>
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<td>0.377</td>
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<td>0.122</td>
<td>1.079</td>
<td>0.284</td>
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<td><strong>Frontal Plane Hip Position</strong></td>
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<tr>
<td>Intercept</td>
<td>3.931</td>
<td>3.310</td>
<td>0.118</td>
<td>1.187</td>
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<td>30.551</td>
<td>-0.038</td>
<td>-0.339</td>
<td>0.736</td>
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<tr>
<td>Peak Gluteus Medius EMG (%MVIC)</td>
<td>0.139</td>
<td>0.624</td>
<td>0.025</td>
<td>0.223</td>
<td>0.824</td>
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References


