Effect of Restricted Hip Flexors on Biomechanical Risk Factors for ACL Injury

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ABSTRACT

MATTHEW MILLS: Effect of Restricted Hip Flexors on Biomechanical Risk Factors for ACL Injury
(Under the direction of Darin Padua)

Anterior Cruciate Ligament injuries are devastating on patients and society. It has been theorized that hip flexor restriction may affect muscles distally in the kinetic chain with no supporting literature. We hypothesized that those with hip flexor restriction will display biomechanics linked to ACL injury. Forty subjects completed functional tasks, range of motion measurements, and strength testing of gluteal muscles. Independent t-tests were run to determine the difference between control subjects and those with restricted hip flexors. Subjects with restricted hip flexors were observed to have less dorsiflexion, and less hip abduction and external rotation. There were no differences in gluteal strength. Restricted subjects were observed to have less anterior pelvic tilt at rest, less gluteus maximus activation during squatting, and exhibited differences in kinematics and kinetics during functional tasks. This indicates hip flexor restriction may lead to biomechanics identified as risk factors for ACL injury, and warrants further examination.

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CHAPTER I

INTRODUCTION

BACKGROUND

Anterior Cruciate Ligament (ACL) injury is an extremely debilitating and commonly occurring injury within athletics worldwide. It is estimated that there are approximately 100,000 to 250,000 per year (Toth and Cordasco 2001; Marshall, Padua et al. 2007), with approximately 1 million physician visits annually in the United States (Marshall, Padua et al. 2007). Most commonly, the ACL is injured in a non-contact mechanism (Griffin, Agel et al. 2000; Agel, Arendt et al. 2005; Hewett T 2011), and frequently during a landing task (Shimokochi and Shultz 2008).

Soccer and basketball are frequently considered to be two major activities that have high risk of ACL injury (Ireland 1999; Agel, Arendt et al. 2005; Faunø P 2006; Yu and Garrett 2007). Soccer is widely thought to be the most popular sport in the world, with an estimated 17 million participates, and 7 million female participants (Lohmander, Ostenberg et al. 2004). Furthermore, soccer was observed to have a higher rate of ACL injury than basketball regardless of gender (Agel, Arendt et al. 2005).

While studies have observed a greater prevalence of ACL injury in males than females, the face-value gender disparity hypothesized to be due to the higher exposure to activity likely to cause ACL pathology (Shea, Pfeiffer et al. 2004; Marshall, Padua et al. 2007). The most commonly injured group was observed to be females ages 16-18 years old (Shea, Pfeiffer et al. 2004), with female soccer players observed to be 2.39 times more likely to sustain injury compared to their male counterparts (Ireland 1999). Furthermore, females were observed to have a prevalence of non-contact ACL injury of 0.17 per contact hour, compared to males rate of 0.05 per contact hour (Ireland 1999). Thus, it is evident that female soccer player are at a high risk of ACL injury, and the aim of research should be to understand and modify risk factors in this population.

ACL injury is devastating on not only the individual patient, but also on society as a whole. ACL injury in Denmark sees $5.4 million dollars billed to insurance companies(Gottlob C. A. 2000; Cumps, Verhagen et al. 2008). The estimated financial burden is approximately $11,500 per patient, which leads to an estimated $1.5 billion dollars in costs per year (Gottlob C. A. 2000). Not only is ACL injury financially devastating, but it is extremely physically disabling to patients. Thirty seven percent of patients never return to the same level of activity (Ardern, Webster et al.), with only 44% returning to competitive sports, and 18% never returning to any sport activity (Ardern, Webster et al.). The mean time from surgery to return to activity is 7.3 months (Ardern, Webster et al.). Furthermore, there is a serious psychological toll placed on those with ACL injury, as those who sustain ACL injury have a higher depression rate for a longer duration than those who suffer concussion (Mainwaring, Hutchison et al. 2010).

ACL injury is also associated with an increased risk of future pathology, as the risk of injury increases between 17% and 33% for a repeated injury, with a 45% increase in risk for meniscus pathology for those who do not undergo surgical repair (Maletius and Messner 1999; Dallalana, Brooks et al. 2007; Pinczewski, Lyman et al. 2007). There is also a significant increase in long-term risk for osteoarthritis (OA)

(Lohmander, Ostenberg et al. 2004; Meunier, Odensten et al. 2007). OA onset occurs 10-20 years earlier in those suffering ACL injury compared to a population with similar demographics, with 50% of patients suffering ACL injury presenting with OA developments (Lohmander, Ostenberg et al. 2004; Meunier, Odensten et al. 2007). In addition, 75% of patients have pain that interferes with their activities of daily life due to OA 12 years following ACL injury (Lohmander, Ostenberg et al. 2004).

While there are many biomechanical risk factors that predispose individuals to ACL pathology, lumbo-pelvic hip complex motion and medial knee displacement have both been linked to ACL injury and mechanism. Ireland identified the "position of no return" in which the hip abductors and extensors "shut down", and fail to effectively control lumbo-pelvic-hip motion. This results in hip internal rotation, hip adduction, knee valgus, and tibial external rotation (Ireland 1999). Knee valgus, and by association, medial knee displacement has frequently been suggested as a biomechanical risk factor for ACL pathology (Hruska 1998; Delp, Hess et al. 1999; Boden, Griffin et al. 2000; Hertel J 2004; Agel, Arendt et al. 2005; Alentorn-Geli, Myer et al. 2009; Chiaia, Maschi et al. 2009). Hip musculature weakness has also been linked to greater anterior pelvic tilt (Popovich and Kulig). Greater anterior pelvic tilt has, in turn, been correlated with ACL injury (Hertel J 2004), and has been theorized to be caused by overactive/tight hip flexors.

Females and males display different biomechanics during athletic activity, which may increase their risk of ACL injury. Females demonstrate greater hip flexion and adduction during a single leg squatting task (Zeller, McCrory et al. 2003), lower gluteal activity during a landing task (Zazulak, Ponce et al. 2005), greater quadriceps/hamstring co-activation ratios (Malinzak, Colby et al. 2001), and range of motion shifting from hip external rotation to hip internal rotation (Brophy, Chiaia et al. 2009). All of the above factors have been theorized to potentially increase ACL injury risk.

Hip flexor restriction has been theorized to affect lumbo-pelvic hip motion, as well as ACL injury risk factors, including medial knee displacement, excessive trunk forward flexion, and femoral internal rotation. Other studies have examined other aspects of hip and trunk biomechanics, and have linked it to ACL pathology risk factors, including lateral trunk motion, overactive hip adductors, and underactive or inhibited hip abductors (Hewett T 2011). Delp et al identified that greater hip flexion is associated with a shifting of the line of pull of the external rotators, which, in turn, forces these muscle to cause an internal rotation moment of the hip (Delp, Hess et al. 1999). Zeller et al identified that overactive rectus femoris musculature during a single leg squat increases knee valgus displacement (Zeller, McCrory et al. 2003). Furthermore, greater flexibility of the hamstrings and lesser activation has been suggested to decrease knee stiffness and increase knee valgus (Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003). Female soccer players possess decreased hip flexor flexibility as measured by a "Thomas test" compared to normative values for a similar demographic that does not play soccer (Chiaia, Maschi et al. 2009). This "tightness" may be a result of either mechanical restriction, or a result of hyperactivity of the hip flexor musculature. As such, further examination of the effects of hip flexor tightness and its relationship to lower extremity biomechanics is indicated.

Therefore, the purpose of this study is to examine the difference in biomechanical risk factors for ACL injury during three dynamic tasks between female soccer players with restricted hip flexors and those with "normal" hip flexor length. Determining this relationship will allow for clinicians to screen for a potential risk factor for ACL pathology, and could potentially lead to prevention strategies through correction of muscle imbalances that characterize hip flexor muscle restriction.

VARIABLES

- *Independent*
  - Hip Flexor Restriction (2 Groups)
    - Control: Those with normal hip flexor motion as measured by a modified "Thomas test" compared to normative values (Figure 1.1)
    - Restricted: Those who display limited hip flexor motion as measured by a modified "Thomas test" compared to normative values (Figure 1.2)
- *Dependent*
  - Kinematic Data
    - Sagittal Plane Hip Kinematics at initial ground contact
    - Frontal Plane Hip Kinematics at initial ground contact
    - Sagittal Plane Knee Kinematics at initial ground contact
    - Frontal Plane Knee Kinematics at initial ground contact
    - Peak Sagittal Plane Hip Kinematics
    - Peak Frontal Plane Hip Kinematics
    - Peak Sagittal Plane Knee Kinematics

        - Peak Frontal Plane Knee Kinematics
        - Peak Trunk Forward Flexion Angle
        - Peak Anterior Pelvic Tilt
        - Sagittal Plane Hip Displacement
        - Frontal Plane Hip Displacement
        - Sagittal Plane Knee Displacement
        - Frontal Plane Knee Displacement
        - Medial Knee Displacement
    - Range of Motion of Lower Extremity Musculature
        - Iliopsoas
        - Rectus Femoris
        - Hamstrings
        - Hip Internal Rotators
        - Hip External Rotators
        - Hip Adductors
        - Gastrocnemius/Soleus
    - Isometric Hip Strength
        - Gluteus Medius
            - Measured at neutral and 20 degrees abduction
        - Gluteus Maximus
            - Measured at neutral and 15 degrees extension
    - Electromyography of Lower Extremity Musculature During Functional Tasks

- Gluteus Maximus
- Biceps Femoris

RESEARCH QUESTION AND RESEARCH HYPOTHESIS

1. *Research Question 1*: Is there a statistically significant difference between groups of female soccer players with restricted hip flexors as measured by a modified "Thomas test" and those with normal hip flexor length in trunk, hip, and knee kinematic data during three functional tasks.
    a. *Research Question 1A*: Is there a statistically significant difference in sagittal plane hip kinematics at initial ground contact, peak values, and displacements between these groups during three functional tasks?
    b. *Research Question 1B*: Is there a statistically significant difference in frontal plane hip kinematics at initial ground contact, peak values, and displacements between these groups during three functional tasks?
    c. *Research Question 1C*: Is there a statistically significant difference in sagittal plane knee kinematics at initial ground contact, peak values, and displacements between these groups during three functional tasks?
    d. *Research Question 1D*: Is there a statistically significant difference in frontal plane knee kinematics at initial ground contact, peak values, and displacements between these groups during three functional tasks?
    e. *Research Question 1E*: Is there a statistically significant difference in medial knee displacement between these groups during three functional tasks?

f. *Research Question 1F*: Is there a statistically significant difference in peak hip extension between these groups during three functional tasks?

g. *Research Question 1G*: Is there a statistically significant difference in peak anterior pelvic tilt between these groups during three functional tasks?

- *Research Hypothesis 1*: The group with restricted hip flexors will have statistically significantly differences in sagittal and frontal plane kinematics at initial ground contact, peak values, and displacements, peak anterior pelvic tilt, peak trunk forward flexion, peak hip extension and medial knee displacement compared to a control group during three functional tasks.
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  - *Research Hypothesis 1A*: The restricted hip flexor group will have significantly more anterior pelvic tilt compared to the control group during three functional tasks.
  - *Research Hypothesis 1B*: The restricted hip flexor group will have significantly more trunk forward flexion compared to the control group during three functional tasks.
  - *Research Hypothesis 1C*: The restricted hip flexor group will have significantly more medial knee displacement compared to the control group during three functional tasks.
  - *Research Hypothesis 1D*: The restricted hip flexor group will have significantly less hip extension compared to the control group during three functional tasks.

2. *Research Question 2*: Is there a statistically significant difference in passive hip range of motion between groups of female soccer players with restricted hip flexors as measured by a modified "Thomas Test" and those with normative hip flexor motion?
    a. *Research Question 2A*: Is there a statistically significant difference in hamstring flexibility between these groups?
    b. *Research Question 2B*: Is there a statistically significant difference in iliopsoas flexibility between these groups?
    c. *Research Question 2C:* Is there a statistically significant difference in rectus femoris flexibility between these groups?
    d. *Research Question 2D:* Is there a statistically significant difference in hip internal rotator flexibility between these groups?
    e. *Research Question 2E:* Is there a statistically significant difference in hip external rotator flexibility between these groups?
    f. *Research Question 2F:* Is there a statistically significant difference in hip adductor flexibility between these groups?
    g. *Research Question 2G:* Is there a statistically significant difference in ankle plantarflexor flexibility between these groups?

- *Research Hypothesis 2*: There will be a statistically significant difference in hip musculature flexibility in a group of female soccer players with restricted hip flexor length and a control group.

- *Research Hypothesis 2A*: The restricted hip flexor length group will have significantly more hamstring flexibility compared to the control group.
- *Research Hypothesis 2B*: The restricted hip flexor length group will have significantly less iliopsoas flexibility compared to the control group.
- *Research Hypothesis 2C*: The restricted hip flexor length group will have significantly less rectus femoris flexibility compared to the control group.
- *Research Hypothesis 2D*: The restricted hip flexor length group will have significantly less hip internal rotator flexibility compared to the control group.
- *Research Hypothesis 2E*: The restricted hip flexor length group will have significantly more hip external rotator flexibility compared to the control group.
- *Research Hypothesis 2F*: The restricted hip flexor length group will have significantly less hip adductor flexibility compared to the control group.
- *Research Hypothesis 2G*: The restricted hip flexor length group will have significantly less ankle plantarflexor flexibility compared to the control group.

3. *Research Question 3*: Is there a statistically significant difference in isometric hip muscle strength between groups of female soccer players with restricted hip

flexor length as measured by a modified “Thomas test” and those with normative hip flexor length?

a. *Research Question 3A*: Is there a statistically significant difference in gluteus maximus isometric strength between these groups at neutral?

b. *Research Question 3B*: Is there a statistically significant difference in gluteus maximus isometric strength between these groups at 15 degrees of hip extension?

c. *Research Question 3C*: Is there a statistically significant difference in gluteus medius isometric strength between these groups at neutral?

d. *Research Question 3D*: Is there a statistically significant difference in gluteus medius isometric strength between these two groups at 20 degrees of hip abduction?

- *Research Hypothesis 3*: There will be a statistically significant difference in hip isometric strength in a group of female soccer players with restricted hip flexor length and a control group.
    - *Research Hypothesis 3A*: The restricted hip flexor length group will have significantly less gluteus maximus isometric strength compared to the control group at neutral.
    - *Research Hypothesis 3B*: The restricted hip flexor length group will have significantly less gluteus maximus isometric strength compared to the control group at 15 degrees of hip extension.

- *Research Hypothesis 3C*: The restricted hip flexor length group will have significantly less gluteus medius isometric strength compared to the control group at neutral.
- *Research Hypothesis 3D*: The restricted hip flexor length group will have significantly less gluteus medius isometric strength compared to the control group at 20 degrees of hip abduction.

4. *Research Question 4*: Is there a statistically significant difference between groups of female soccer players with restricted hip flexor length as measured by a modified "Thomas test" and those with normal hip flexor length in trunk, hip, and knee kinetic data during three functional tasks.
    a. *Research Question 4A*: Is there a statistically significant difference in peak hip extension moment between these groups during three functional tasks?
    b. *Research Question 4B*: Is there a statistically significant difference in peak knee extension moment between these groups during three functional tasks?
    c. *Research Question 4C*: Is there a statistically significant difference in peak hip adduction moment between these groups during three functional tasks?
    d. *Research Question 4D*: Is there a statistically significant difference in peak knee varus moment between these groups during three functional tasks?

- *Research Hypothesis 4*: The group with restricted hip flexor length will have statistically significantly differences in peak anterior pelvic tilt, peak trunk forward flexion, peak hip extension and medial knee displacement compared to a control group during three functional tasks.
    - *Research Hypothesis 4A*: The restricted hip flexor length group will have significantly less peak hip extension moment compared to the control group during three functional tasks.
    - *Research Hypothesis 4B*: The restricted hip flexor length group will have significantly more knee extension moment compared to the control group during three functional tasks.
    - *Research Hypothesis 4C*: The restricted hip flexor length group will have significantly more peak hip adduction moment compared to the control group during three functional tasks.
    - *Research Hypothesis 4D*: The restricted hip flexor length group will have significantly more peak knee varus moment compared to the control group during three functional tasks.
- *Research Question 5*: Is there a statistically significant difference in groups of female soccer players with restricted hip flexor length as measured by a modified "Thomas test" and those with normal hip flexor length in average hip extensor electromyographic activity during three functional tasks?
    - *Research Question 5*A: Is there a statistically significant difference in average Gluteus Maximus activation between these groups during three functional tasks?

- *Research Question 5B:* Is there a statistically significant difference in average Biceps Femoris activation between these groups during three functional tasks?

- *Research Hypothesis 5:* The group with restricted hip flexor length will have statistically significantly differences in average electromyography of hip extensors compared to a control group during three functional tasks.
  - *Research Hypothesis 5A*: The restricted hip flexor length group will have significantly less average Gluteus Maximus activation during three functional tasks.
  - *Research Hypothesis 5B*: The restricted hip flexor length group will have significantly less average Biceps Femoris activation compared to the control group during three functional tasks.

Statistical Hypothesis

- H_O: HFT=CG
- H_A: HFT≠CG

OPERATIONAL DEFINITIONS

- Hip Flexor Muscles
  - Iliopsoas and Rectus Femoris as measured by a Modified "Thomas Test" (Harvey 1998)
- Modified "Thomas Test":
  - A hip flexor length test performed with the subject lying supine on a table with the knee to be tested flexed to 90 degrees and the subject holds the

contralateral leg into maximal hip flexion. The examiner then lowers the braced leg until the pelvis begins to anteriorly rotate or the first point of resistance. The angle is measured between the table and the thigh and recorded (Ferber, Kendall et al. 2010).

- Restricted Hip Flexor Tightness
    - Subjects, when measured using a modified “Thomas test” have digital inclinometer readings of the ipsilateral thigh of greater than 0 degrees above the horizontal (Ferber, Kendall et al. 2010)
- Non-Contact Injury
    - ACL failure (rupture) during a functional activity in the absence of any external force except ground reaction forces (Shimokochi and Shultz 2008)
- Lumbo-pelvic hip complex motion
    - Any physiologic movement that occurs in the Lumbar Spine, Femoro-acetabular joint, Sacroilliac joint, or rotation of the innominate.
- Anterior Pelvic Tilt
    - Motion (in degrees) of the ASIS in an anterior direction of the sagittal plane as measured through motion analysis.
- Forward Trunk Flexion
    - Motion (in degrees) of the trunk rigid body segment angle relative to the world in the sagittal plane as measured through motion analysis.
- Medial Knee Displacement

    - Movement of the joint center of the knee medially in the frontal plane over the great toe. (Bell, Padua et al. 2008)
- Female Soccer Player
    - A player who participates in organized soccer activities at least 2 times per week for at least 45 minutes per occasion who plays under standard rules and regulations.
- Jump Landing Task
    - A procedure in which the participant jumps forward off a 30-cm high box placed at a distance ½ of their body height from the leading edge of a right and left force plate. The participant is instructed to land with their right foot on the right force plate and their left foot on the left force plate. The participant lands and is instructed to immediately jump for maximal vertical height.
- Single Leg Squat Task
    - A task in which a subject, while standing on their dominant leg, lowers their center of mass until the thigh is parallel to the ground, or the subject loses their balance and touches down with their unaffected side.
- Dominant Leg
    - The leg in which the subject would voluntarily choose to kick a soccer ball for maximal distance.(Zeller, McCrory et al. 2003)

ASSUMPTIONS

- A handheld dynamometer accurately depicts isometric strength at a specific joint angle for a specific muscle
- Motion tracking is representative of physiologic motion compared to skin motion with minimal motion artifact
- The use of a standard goniometer to measure active range of motion is indicative of the antagonist's length.
- The use of a digital inclinometer to measure active range of motion is indicative of the antagonist's length.
- EMG electrodes on the skin may not give a true reading of the underlying muscle activity

DELIMITATIONS:

- Subjects will be Female Soccer Players from the University of North Carolina at Chapel Hill Women's Varsity, Club, Intramural, and Lifetime Fittness Soccer Programs
- Subjects will be ages 18 to 35 years old
- Subjects with history of lower extremity, abdominal, or spine injury in the past 3 months that limited activity for greater than three days will be excluded.
- Subjects with a history of any lower extremity, abdominal, or spine fracture or surgery will be excluded.

LIMITATIONS:

- Selected athletes may not accurately represent all soccer players
- Soccer players may not accurately represent other field sports
- Different subjects may have previous training in performing specified tasks
- Laboratory environment may not accurately represent field environment
- Motion artifact may limit the accuracy of motion capture devices.
- Footwear is not consistent among individuals and thus may lead to alterations
    - Tennis Shoes may not affect the LE biomechanics in the same manner as soccer cleats during activity.
- This study did not measure the muscular activation of the other hip, thigh, and lower leg muscles.

SIGNIFICANCE OF THIS STUDY

The identification of ACL risk factors is of critical importance due to the devastating nature of these injuries. While many studies have identified biomechanical and anatomic risk factors at the hip and knee, very little research has examined a mechanistic connection between the hip flexors and ACL injury. While the theory of reciprocal inhibition and synergistic dominance affecting the knee has been proposed, there is no literature substantiating that theory. This study will provide insight regarding the connections between the hip flexors and ACL risk factors, and can help with the identification of at risk individuals and prevention of ACL injury.

CHAPTER II

REVIEW OF THE LITERATURE

INTRODUCTION

Anterior Cruciate Ligament (ACL) injuries are debilitating, devastating, and costly injuries for patients and the healthcare system. Previous literature has looked at a wide array of various risk factors. These risk factors have included anatomical risk factors, as well as biomechanical risk factors. While anatomical risk factors are widely considered to be non-modifiable, biomechanical risk factors have been shown to be modifiable frequently (Hewett, Stroupe et al. 1996). Biomechanical risk factors have been identified proximally at the region of the lumbo-pelvic-hip complex (Ireland 1999; Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003; Hewett T 2011), at the knee itself through kinetics and kinematics (Hruska 1998; Delp, Hess et al. 1999; Boden, Griffin et al. 2000; Hertel J 2004; Agel, Arendt et al. 2005; Alentorn-Geli, Myer et al. 2009; Chiaia, Maschi et al. 2009), as well as distally at the foot (Hertel J 2004).

A common finding to both the proximal and distal biomechanical risk factors are that they all have been theorized to affect the body in a way that leads to the "position of no return" as described by Ireland et al (Ireland 1999). This position has been linked to ACL injury, and thus is considered to be a lower extremity posture associated with injury. One potential risk that has not been examined, however, is hip flexor tightness. While it has been theorized that hip flexor tightness can cause a negative cascade distally in the kinetic chain, no methodology to date has substantiated this theory. Therefore, the purpose of this literature review is to examine the relationship between restricted hip flexor tightness and previously identified lower extremity biomechanical risk factors for ACL injury.

DEFINITION OF ACL INJURY

An ACL tear is commonly defined as a rupture of the Anterior Cruciate Ligament, an intracapsular ligament within the knee responsible for preventing anterior translation of the tibia on the femur (Neuman 2010). The ACL has two distinct bundles, the anterior-medial bundle, as well as the posterior-lateral bundle, which are named based on their relative attachments on the tibia (Neuman 2010). It is believed that the ACL provides approximately 85% of total passive resistance of anterior translation of the tibia(Neuman 2010). Tension within the cruciate ligaments in the knee is also responsible for assisting with the arthrokinematics of the knee, as well as proprioceptive feedback of the knee. Tearing of the anterior cruciate ligament, therefore, presents major complications, primarily knee instability (Noyes, Mooar et al. 1983).

ACL injury is commonly classified by the literature as either contact or non-contact. A contact injury is defined as an injury in which the patient injured the ACL due to external forces besides simply ground reaction forces. A non-contact injury is defined as an injury in which the patient injured the ACL during functional activities in the absence of any external forces other than the ground reaction force (Shimokochi and Shultz 2008).

Approximately 70% of ACL injuries are non-contact in nature (Agel, Arendt et al. 2005), suggesting that an individual's movement itself was the cause of injury, thus, non-contact ACL injuries can be prevented through alterations of the athlete's biomechanics. In cadaveric studies, the ACL was observed to be torn with anterior tibial shear forces, which results in anterior tibial translation, as well as increased stress on the ACL with valgus and knee flexion (Kennedy, Weinberg et al. 1974; Withrow, Huston et al. 2006). In human subjects, there were observed to be three main positions of risk. It was observed that the greatest risk in a functional movement was knee internal rotation in full knee extension, with another position during varus loads in knee hyperextension and extension (Markolf, Burchfield et al. 1995). Lastly, valgus loads in knee flexion were observed to lead to increased risk of injury (Markolf, Burchfield et al. 1995). As such, the vast majority of ACL injuries may be preventable, which requires a better understanding of the underlying risk factors and how they may predispose an athlete for these devastating injuries, as well as a better way to screen for these risk factors.

ANTERIOR CRUCIATE LIGAMENT INJURY PREVALENCE

ACL injury is a commonly occurring athletic injury, with frequency estimates ranging from 100,000 to 250,000 injuries occurring per year in the United States (Toth and Cordasco 2001; Marshall, Padua et al. 2007). Furthermore, there are approximately 1 million physician visits annually in the United States for treatment of cruciate ligament injury. (Marshall, Padua et al. 2007). These injuries are most likely to occur during activities with high impact and high velocity motion, including jumping, landing and cutting. (Boden, Griffin et al. 2000; Toth and Cordasco 2001; Alentorn-Geli, Myer et al. 2009; Boden, Torg et al. 2009; Hewett, Torg et al. 2009).

Activities with rapid accelerations and changes in direction, as described above, are common during athletic participation. One such sport that is widespread throughout the globe is soccer, in which athletes are required to have high velocity movements and changes in direction. It is currently estimated that there are 17 million players, with 7 million female athletes (Lohmander, Ostenberg et al. 2004). As such, soccer is considered to be a sport that places participants at high risk for ACL injury. This is validated through the finding that athletes were observed to have the highest rate of injury in soccer compared to basketball per contact hour across genders (Agel, Arendt et al. 2005).

The prevalence of ACL injury in soccer athletes has been repeatedly examined, and it was observed that there was the highest prevalence of injury in male athletes, yet the highest risk of injury in female athletes (Shea, Pfeiffer et al. 2004; Marshall, Padua et al. 2007). The higher prevalence has been attributed to the fact that they have higher exposure to activities likely to cause ACL pathology (Shea, Pfeiffer et al. 2004; Marshall, Padua et al. 2007). Further analysis of epidemiological data suggests there is a higher incidence of injury in females (Griffin, Agel et al. 2000; Shea, Pfeiffer et al. 2004), particularly those ages 16-18 (Shea, Pfeiffer et al. 2004). It was observed that female soccer players are 2.29 times more likely to sustain than male counterparts (Ireland 1999), as they have noncontact injury rates of 0.17 per contact hour in females, compared to 0.05 per contact hour in males (Ireland 1999). This finding was confirmed by Uhorchak (2003), who observed that females are 3x more likely than males to sustain ACL injury (Uhorchak, Scoville et al. 2003). As such, it is clear that there is a gender bias among those who suffer ACL injury, particularly within soccer athletes, which mandates that research examine potential risk factors in females.

NEGATIVE EFFECTS OF ACL INJURY

ACL injuries are also extremely debilitating for patients physically and emotionally. It was observed that 37% of patients never return to same level of competition (Ardern, Webster et al.). In addition, only 44% return to competitive sports and 18% never return to sports at all (Ardern, Webster et al.). The mean time from surgery until the athlete returns to athletic activity was observed to be 7.3 months (Ardern, Webster et al.). Furthermore, it was observed that there was a greater level and duration of depression compared to those suffering from a concussion (Mainwaring, Hutchison et al. 2010).

There is also high risk of a second ACL tear, as after a patient tears their ACL, individuals with a previous ACL injury were observed to be at an increased risk of repeated injury, either on the affected side or contralateral side. It was observed that the risk of re-injury ranged from 33% (Dallalana, Brooks et al. 2007) to 27.2% (Pinczewski, Lyman et al. 2007). Furthermore, it was observed that there was an increase in risk for meniscus injury, as it was observed that there was a 45% risk for meniscus injury (Maletius and Messner 1999).

It was also observed that ACL injury increases the long-term risk for osteoarthritis within patients. It was observed that osteoarthritis onset began 10-20 years earlier than a non-injured population of similar demographics (Meunier, Odensten et al. 2007).

Furthermore, 10 to 20 years after diagnosis, 50% of patients were observed to have osteoarthritic development (Lohmander, Englund et al. 2007), with 75% of patients reporting pain that interferes with daily life due to osteoarthritis 12 years after ACL injury (Lohmander, Ostenberg et al. 2004).

ACL injuries are not only devastating on the individual, but also financially demanding on the healthcare system. It was observed that $5.4 million billed to insurance companies per year with $2.7 million covered by government in Denmark (Cumps, Verhagen et al. 2008). Furthermore, it is estimated that the average cost of ACL injury was $11,500 per patient (Gottlob C. A. 2000), which leads to an estimated cost of $1.5 billion per year (Gottlob C. A. 2000).

RISK FACTORS FOR ACL INJURY

Anatomical

While there are many modifiable risk factors for ACL injury, there are also several factors that are based on the subject's anatomy, and thus are considered to be non-modifiable. These factors can occur both proximally and distally within the kinetic chain. One prominent anatomic risk factor proximally in the kinetic chain is the Quadriceps Angle, also known as the "Q-angle". The Q angle is defined as the angle formed with a straight line from the anterior superior iliac spine to the patellar center with the line running from the patellar center to the tibial tuberosity (Daneshmandi, Saki et al. 2011). Females have been observed to have larger values of the Q angle compared to their male counterparts, which was hypothesized to be a result of wider hips (Hertel J 2004; Pantano, White et al. 2005). Several studies have linked increased Q angle to increased risk of ACL injury (Hertel J 2004; Pantano, White et al. 2005; Tillman, Bauer et al. 2005; Daneshmandi, Saki et al. 2011). Furthermore, increased Q angle has been linked with increased knee valgus angulation, which is the most commonly referenced risk factor for ACL injury (Powers 2003). With respect to the distal kinetic chain, foot pronation has also been linked to ACL injury risk (Hertel J 2004). While foot pronation can be modified through the use of orthotics, Q-angle is not considered to be a modifiable risk factor, and is frequently larger in females compared to males due to their wider pelvis, and is considered to be a potential contributor to the increased risk of ACL injury for females (Boden, Griffin et al. 2000; Hewett, Myer et al. 2006).

While anatomical factors proximally and distally in the kinetic chain have been linked with increased ACL risk, knee anatomical risk factors have consistently been linked to increased risk. One such risk factor that was identified was a decreased intercondyler notch size, which is frequently observed more in females than males (Boden, Griffin et al. 2000; McClay Davis and Ireland 2001; Tillman, Smith et al. 2002; Uhorchak, Scoville et al. 2003; Alentorn-Geli, Myer et al. 2009; Boden, Torg et al. 2009). ACL biomechanics have also been linked to ACL injury, as the specific biomechanical properties vary depending upon gender (Chandrashekar, Slauterbeck et al. 2005; Chandrashekar, Mansouri et al. 2006). Furthermore, a decreased ACL size has been linked to ACL injury, and females have typically a smaller ACL compared to their male counterparts (Chandrashekar, Slauterbeck et al. 2005; Chandrashekar, Mansouri et al. 2006; Hashemi, Mansouri et al. 2011).

An area of specific controversy, specifically when addressing gender specific risk factors includes the presence and concentration of specific hormones, predominantly estrogen and progesterone. Several studies have observed that decreases in estrogen and progesterone in women have been linked to increased ligament laxity, and thus ACL injury (Boden, Griffin et al. 2000; Slauterbeck and Hardy 2001; Toth and Cordasco 2001; Slauterbeck, Fuzie et al. 2002; Wojtys, Huston et al. 2002). This is also substantiated as ACL injury frequencies are not uniform throughout menstrual cycle (Slauterbeck, Fuzie et al. 2002; Wojtys, Huston et al. 2002). However, other studies have refuted this claim, and observed no link between ACL rupture and hormone concentrations in females (Warden, Saxon et al. 2006).

Muscle Function

Various muscular inhibitions and lengthened muscles have also been linked to ACL injury. Brophy et al observed that decreased hip external rotation range of motion may be correlated to ACL injury risk, and is also more commonly observed in females compared to males (Brophy, Chiaia et al. 2009). On the contrary, Bell et al (2008) demonstrated that increased external rotation range of motion in the individuals with excessive medial knee displacement during a squatting task (Bell, Padua et al. 2008). Furthermore, hamstring hypermobility has been linked to increased ACL risk (Boden, Dean et al. 2000) as it decreases dynamic control of the knee (Hewett, Stroupe et al. 1996; Huston and Wojtys 1996), and would limit its function as an ACL agonist. Males were observed to have tighter hamstrings on average compared to females of similar demographics (Krivickas and Feinberg 1996). This may indicate that males have a decreased risk of ACL injury due to increased dynamic control of the knee with hyperactivity and associated hypomobility of the hamstrings.

Decreased production of hamstring force has also been linked with ACL pathology (More, Karras et al. 1993; Shimokochi and Shultz 2008; Alentorn-Geli, Myer et al. 2009). Hamstrings have been long considered to be vital in assisting the ACL in preventing anterior tibial translation and stabilizing the knee (Solomonow, Baratta et al. 1987; Boden, Griffin et al. 2000; Kwak, Ahmad et al. 2000). As such, muscle inhibitions have been observed to increase ACL risk factors. There has also been speculation regarding overactive hip flexors affecting the hamstrings, however, there is no current literature supporting that hypothesis.

Muscle strength or hyperactivity has also been linked to ACL injury risk. Most commonly associated with ACL risk is a large unopposed quadriceps force (Boden, Dean et al. 2000; Boden, Griffin et al. 2000; Yu and Garrett 2007; Shimokochi and Shultz 2008; Kulas, Hortobagyi et al. 2010), as that provides an anterior translation of the tibia due to the attachment through the patellar tendon onto the anterior tibia via the tibial tuberosity. This was validated through the finding of increased reliance on the quadriceps for tibial stabilization in females compared to hamstrings in males (Huston and Wojtys 1996). The use of the quadriceps for stabilization may lead to increased anterior tibial shear force compared to male counterparts.

Greater hamstring stiffness has also been linked to decreased ACL injury risk, as Blackburn et al (2011) also observed decreased anterior tibial translation with increased hamstring stiffness (Blackburn, Norcross et al. 2011). Furthermore, subjects exhibiting greater hamstring stiffness display better landing biomechanics through improved dynamic control of the knee (Blackburn, Bell et al. 2009). There also appears to be a gender bias, as hamstring stiffness has been observed to be higher in males compared females (Blackburn, Padua et al. 2004; Blackburn, Bell et al. 2009). Furthermore, hamstring neuromechanics were observed to be limited in females (Blackburn, Bell et al. 2009), as the rate of force production and time to 50% force were found to be slower in females compared to their male counterparts. This was correlated with decreased hamstring stiffness (Blackburn, Bell et al. 2009), which was linked to ACL injury risk. As such, determining potential causes of decreased hamstring stiffness, including reciprocal inhibition from hip flexor tightness, should be a point of emphasis for future research.

Functional Task Biomechanics

Another method for identification of risk factors includes functional screening through examination of form and techniques in specific tasks. Within landing tasks, there have been several compensations and motions that have been identified as predisposing factors for ACL injury. One identified factor was increased hip flexion and hip abduction at initial ground contact, as these motion patterns seem to increase loading on the ACL and thus would predispose the athlete for injury (Boden, Torg et al. 2009). This was also observed by Krosshaug et al (2007) in which they found that females had increased hip flexion while sustaining ACL injury compared to their male counterparts (Krosshaug, Nakamae et al. 2007). Furthermore, Zazulak et al (2005) hypothesized that increased hip flexion reduced the body's ability to absorb the weight of the upper extremity, which would lead to and increase in ACL injury (Zazulak, Ponce et al. 2005). However, causes for increased hip flexion have been not examined, but it has been theorized that increased tightness or over-activity of the hip flexor group would lead to an increase in hip flexion at ground contact.

A second identified risk factor is landing with the knee flexed less than 30 degrees at initial ground contact (Boden, Dean et al. 2000; Lephart, Abt et al. 2002; Decker, Torry et al. 2003; Padua, Marshall et al. 2004; Salci, Kentel et al. 2004; Warden, Saxon et al. 2006). Landing with decreased knee flexion also increases the vertical ground reaction forces, which increases the load placed on the ACL, which accordingly would increase the risk of ACL injury (Nigg 1985). This compensation could be affected by the hip flexor group, due to the fact that over-activity and tightness of rectus femoris could cause decreased knee flexion through its attachment on the anterior tibia via the patellar tendon.

The most commonly examined ACL risk factor during functional activity, however, is knee valgus, or medial knee collapse (Hewett, Stroupe et al. 1996; Padua, Marshall et al. 2004; Shimokochi and Shultz 2008). It was observed that knee valgus was associated with a 30% increase in ACL loading compared to loading with flexion alone in cadaveric knees (Withrow, Huston et al. 2006). It has also been observed that females exhibit greater medial knee collapse than males while performing a landing task (Ford, Myer et al. 2003), as well as performing a jump landing (Padua, Marshall et al. 2004). Medial knee collapse has been linked in theory to the cascade of effects from hip flexor tightness, as inhibition of the hamstrings and gluteus maximus would lead to use of the hip adductors as an extension mechanism. This would lead to excess medial knee displacement, as Mauntel et al (2011) observed that individuals with medial knee displacement during a single leg squatting task display greater hip adductor activity relative to gluteus maximus compared to control subjects (Mauntel, Padua et al. 2011).

Medial knee displacement is a likely cause of ACL injury due to causing impingement of the ACL by the lateral femoral condyle, which leads to traumatic shearing and thus rupture (Kennedy, Weinberg et al. 1974).

However, knee valgus has been linked to other motions proximally in the kinetic chain. Hollis et. al (1991) first noted that knee valgus was linked to hip internal rotation as weight bearing knee flexion angles increase during a squatting task (Hollis, Takai et al. 1991). Shin et. al (2011) also observed that knee valgus in association with hip internal rotation increased the strain on the ACL more than either motion individually (Shin, Chaudhari et al. 2011). McLean et al (2005) also replicated these findings during a sidestepping task, as they observed that knee valgus was associated with increased hip flexion and femoral internal rotation (McLean, Huang et al. 2005), both of which have been theorized to be affected by the hip flexor group, as well as compensatory motions of synergistic muscles.

There also appears to be findings regarding hip internal rotation, which was correlated with functional knee valgus, and it's relevance to ACL injury. Female soccer players were observed to have greater femoral internal rotation compared to normative values of similar demographics (Chiaia, Maschi et al. 2009), as well as when compared to males of similar demographics (Brophy, Chiaia et al. 2009). Femoral internal rotation has also been observed to be a risk factor for ACL pathology when combined with other rotational movements (Shimokochi and Shultz 2008; Alentorn-Geli, Myer et al. 2009).

Another functional risk factor for ACL pathology was observed to be an increase in tibial internal rotation (Padua, Marshall et al. 2004; Alentorn-Geli, Myer et al. 2009). However, this is in contrast to the most commonly referenced position of ACL risk, which was defined by Ireland (1999), where she described the “position of no return”, a pattern in which the hip abductors and extensors “shut down”, which results in hip internal rotation, hip adduction, knee valgus, and tibial external rotation. This position is referred to as “no return”, as it is considered to be an extremely high risk for ACL injury (Ireland 1999).

This “position of no return” which Ireland described could be caused through a cascade of events stemming from the hip flexor group, as overactive hip flexors could result in the “shutting down” of the gluteal group and hamstrings, which Ireland described as the hip extensors and abductors. As such, as Ireland described, it could lead to this position of extremely heightened risk for injury, and thus requires further examination.

LUMBO-PELVIC HIP COMPLEX CONTRIBUTIONS

Anatomy of Hip Flexor Group

The primary hip flexor in the human body is considered to be the Illiopsoas muscle group, responsible for hip flexion, anterior tilting of the pelvis, as well as trunk flexion. It has also been observed to assist with hip external rotation with the hip abducted, and can assist with vertical stability of the spine (Neuman 2010). The muscle is divided into three portions. The first is Illiacus which runs from the iliac fossa and lateral edge of sacrum and runs to the lesser trochanter (Neuman 2010). The second aspect is the Psoas Major, which runs from the transverse process of the lumbar vertebrae and attaches on the lesser trochanter of the femur (Neuman 2010). The third, and most commonly forgotten aspect is Psoas Minor, which runs from between twelfth thoracic and first lumbar vertebra and attaches to the pelvis near pectinial line. Unlike the other aspects, it is responsible for posterior pelvic tilting, but has been observed to be missing in up to 40% of people (Neuman 2010).

Another muscle that is primarily responsible for hip flexion is Rectus Femoris. It originates from the anterior inferior iliac spine and the superior rim of the acetabulum to the tibia via the patellar tendon. Its actions are acting as the primary knee extensor, as well as providing approximately 1/3 of the total isometric hip flexion torque (Neuman 2010)

There are several other muscles that assist with hip flexion, including Sartorius, Tensor Fascia Latte, Adductor Longus, and Pectineus. Sartorius originates on the anterior superior iliac spine, and inserts on the medial surface of the proximal tibia. It also is responsible for hip abduction and external rotation, which places the individual in the "figure four" position (Neuman 2010). Tensor Fascia Latte originates on the illium just lateral to Sartorius, and attaches on the proximal band of the Iliotibial Band, which extends to the lateral tubercle of the tibia. It is also responsible for hip abduction, hip internal rotation, and assists with stabilizing the lateral aspect of the knee (Neuman 2010)

The Adductor Longus originates on the superior and inferior pubic rami and the adjacent body of pubis, and inserts on the middle 1/3 of the medial lip of the linea aspera on the femur. It is responsible for hip adduction, as well as hip flexion when the thigh is in less than 40 degrees of hip flexion, as well as hip internal rotation (Neuman 2010). Lastly, Pectineus originates on the superior and inferior pubic rami and adjacent body of pubis, and inserts on the linea aspera of the femur. It is also responsible for hip adduction and internal rotation along with hip flexion (Neuman 2010).

Anterior Pelvic Tilt

Along with providing the body with hip flexion, the hip flexor group also has been associated with assisting in anterior pelvic tilting. In particular, the hip flexors provide a strong anterior pelvic tilt unless the rectus abdominis is able to provide a counteracting posterior pelvic force strong enough to counterbalance that action (Hodges and Richardson 1997; Neuman 2010). Furthermore, anterior pelvic tilt has been linked to ACL pathology (Delp, Hess et al. 1999; Hertel J 2004; Alentorn-Geli, Myer et al. 2009; Chiaia, Maschi et al. 2009), as it can create hip internal rotation, which places the ACL in an increased risk position (Brophy, Chiaia et al. 2009). Furthermore, it was observed that hip weakness may lead to increased anterior pelvic tilt (Popovich and Kulig). As such, hip flexor tightness has been theorized as one potential cause for more hip flexion and anterior pelvic tilting in individuals, which could position the hip and pelvis in positions that may compromise the ACL.

Effects of Hip Flexor Tightness

Flexibility of hip flexors has been theorized to affect a wide array of biomechanical abnormalities, and pathologic conditions. Winters et. al (2004) observed that tightness of the hip flexors can lead to decreased hip extension motion (Winters, Blake et al. 2004). This was linked to anterior pelvic tilting by Schache et. al (2000), as hip extension deficits were observed to lead to anterior pelvic tilting, which had previously been established as a risk factor for ACL injury (Schache, Blanch et al. 2000). Furthremore, hip flexor tightness has been linked to an increased risk for low back pain (Kolber and Fiebert 2005), as well as increased injury incidence (Krivickas and Feinberg 1996).

Tightness in the hip flexor group has also been shown to affect the hamstrings (Chumanov, Heiderscheit et al. 2007; Riley, Franz et al. 2010) as it was linked to an increase in incidence of hamstring strains (Gabbe, Bennell et al. 2006). This is vital to ACL injury prevention, as Withrow et al (2008) observed that in cadaveric knees, increased hamstring force was associated with a greater than 70% reduction of force placed on the ACL (Withrow, Huston et al. 2008). Since the hamstrings are a synergist of the ACL, the effects of hip flexor tightness on the hamstrings could be detrimental to the stability of the knee and the ACL.

SUMMARY

ACL injury is a common and devastating problem not only for the individual who sustains the injury, but also for society as a whole. It has negative effects physically, psychologically, and socioeconomically, which mandates the need for identifying modifiable risk factors to decrease the incidence of these injuries. There are many previously established biomechanical risk factors that have been observed to play a role in increasing risk of ACL injury. Many of these previously established risk factors have also been linked to hip flexor tightness, and hip flexor tightness has also been shown to affect many other negative conditions of the body. As such, it is essential to identify if hip flexor tightness can be added to the base of literature regarding risk factors for ACL injury, and should be targeted in a comprehensive rehabilitation program, and thus, potentially decrease the overall risk of non-contact ACL injury, and thus prevent the associated physiological and psychological effects.

CHAPTER III

METHODOLOGY

SUBJECTS

A total of 40 females were selected from the women's lifetime fitness, intramural, club, and varsity soccer teams at the University of North Carolina at Chapel Hill. The restricted hip flexor group consisted of 20 females, while the control group consisted of 20 females with no difference between groups for height, weight, or age. Group sizes were based on power calculations for an estimated power calculation of 0.80 based on effect size from previous studies (Ford, Myer et al. 2003; McLean, Huang et al. 2005; Pantano, White et al. 2005).

Each participant was assigned to either the "normal" group, or the "restricted" group based on her score of a modified Thomas test during a preseason screening. The modified Thomas test has been shown to have good reliability (Gabbe, Bennell et al. 2004; Peeler and Anderson 2007; Clapis, Davis et al. 2008).

Inclusionary Criteria

All subjects were female soccer players from the University of North Carolina at Chapel Hill, which was defined as, "A player who participates in organized soccer activities at least 2 times per week for at least 45 minutes per occasion." Subjects were between the ages of 18 and 35 years.

Exclusionary Criteria:

Participants were excluded from this study if they had any lower extremity, spine, or abdominal injury in the last 3 months that limited them for greater than 3 consecutive days. Participants with any lower extremity surgery or fracture were also excluded. Participants were also excluded if they had any current vestibular or mild traumatic brain injury. Furthermore, participants who fell between horizontal and 15 degrees below horizontal were excluded.

INSTRUMENTATION

Electromagnetic Motion Capture System

A TrackStar (Ascension Technologies, Inc, Burlington, VT) electromagnetic motion tracking system was used to track lower extremity kinematics. The device consists of an extended range transmitter that emits an electromagnetic field and standard receivers (dimensions 25.4 X 25.4 X 20.3 mm) that detect the electromagnetic field. The TrackStar System tracked and recorded the positions and orientation of the receivers about the x, y, and z axes relative to the transmitter. The device was used to sample lower extremity kinematics at 140 Hz. Electromagnetic tracking systems have been observed to be reliable (An, Jacobsen et al. 1988), and accurate (An, Jacobsen et al. 1988; Milne, Chess et al. 1996) for three dimensional movement of body segments and joints in kinematic analysis.

Digital Inclinometer

Joint angles for measures of flexibility of the rectus femoris, iliopsoas, hip internal rotators, hip external rotators, ankle plantarflexors, and hamstrings were measured using a digital inclinometer (Saunders Group, Inc., Chaska, MN). Intersession and intrarater reliability of the active range of motion testing procedure of the investigator responsible for taking the measures in this study were calculated with intraclass coefficients (ICC) and standard errors of the measurement (SEM) for each range of motion measurement (ICC 3,k range 0.996-0.965, SEM Range 0.2502-2.024) (Table 2). A digital inclinometer has been found to be reliable and valid for these measurements (Bierma-Zeinstra, Bohnen et al. 1998).

Standard Goniometer:

Joint angles for measures of flexibility of the hip adductors and plantar flexors were measured using a standard 30.5 cm (12 in) plastic goniometer. Intersession and intrarater reliability of the passive range of motion testing procedure of the investigator responsible for taking the measures in this study was calculated with intraclass coefficients (ICC) and standard error of the measurement (SEM) for each range of motion measurement (ICC 3,k range, .909-.992; SEM range, 0.836-1.099) (Table 3.2).

Isometric Dynamometer:

A handheld digital dynamometer (Chatillon MSC-500, AMETEK, Inc., Largo, FL) was used to collect peak and mean isometric strength values for the gluteus maximus and gluteus medius. It has been found to be reliable for measurement of hip extension strength (ICC 0.75-0.85) (van der Linden, Aitchison et al. 2004), as well as valid for these measurement (Reed, Den Hartog et al. 1993; Trudelle-Jackson, Jackson et al. 1994). Intersession and intrarater reliability of the isometric strength testing were calculated using intraclass coefficients (ICC) and standard error of the measurement (SEM) for each muscle tested (ICC 3,k range 0.862-0.979, SEM Range 0.477-1.445) (Table 3).

Force Plate

A non-conductive force plate (Bertec 4060-NC, Columbus, OH) was used to sample ground reaction force data at 1000 Hz during the jump-landing and squatting tasks.

Electromyography

A surface electromyography (EMG) system (Bagnoli-8; Delsys, Inc, Boston, MA) with an interelectrode distance of 10 mm, amplification factor of 1,000 (20 – 45 Hz), common-mode rejection ratio of 60 Hz (>80 dB), and input impedance > 1015//0.2 Ω//pF was used to record lower extremity muscle activity. Kollmitzer et al. (Kollmitzer, Ebenbichler et al. 1999) showed EMG measures of lower extremity muscle activity to be reliable for short-term and long-term test-retest intervals.

PROCEDURES

Screening Session

Prior to data collection, each participant underwent a screening process to determine her inclusion into the study and group assignment. The participants read and signed an informed consent form approved by the Institutional Review Board (IRB) of the University of North Carolina at Chapel Hill, and each participant was able to ask questions to clarify any part of the informed consent form prior to signing it. All participants also completed a questionnaire to confirm inclusion/exclusion criteria, the participant's dominant leg (the leg that they would use to kick a soccer ball for maximum distance), medical history, and contact information.

Participants underwent a screening process to determine group assignment. The screening protocol consisted of each participant completing a modified Thomas test and was performed with the subject lying supine on a table with both legs held tight to chest. The examiner stabilized the low back, sacrum, and pelvis, slowly lowering the participant's leg to the point of first resistance. A digital inclinometer was placed along anterior aspect of thigh between ASIS and patella halfway from the superior pole of patella to ASIS in order to determine hip flexor "tightness".

The average of the two trials was recorded and subjects were classified based on normative values of greater than or equal to 0° above the horizontal axis for "restricted", and greater than or equal to 15° below horizontal for "control"(Ferber, Kendall et al. 2010). Subjects who fell between the two classifications were disqualified. Selected participants were contacted at a later date to complete data collection.

On the day of data collection, previously selected subjects reported to the Sports Medicine Research Laboratory on one occasion for testing. Participants' height and weight were recorded. Participants then completed a warm-up on a stationary cycle ergometer at a self-selected pace for five minutes at a rate of perceived exertion of 3/10.

Range of Motion Measurements

All lower extremity range of motion measurements were measured with a digital inclinometer or standard goniometer. Intersession and intrarater reliability and precision were established prior to data collection (Table 2). For each of the following muscle groups, the examiner moved the participant until point of first resistance or compensation from accessory motion is noted. Two trials were taken for each range of motion measurement and the arithmetic mean was calculated for each movement. The following procedures were used:

- **Iliopsoas**: The participant was positioned prone with her knee fully extended with the hips stabilized to the table using webbing. The contralateral leg was stabilized to the table using webbing. The subject was instructed to relax, and the examiner moved the thigh posteriorly until the point of first resistance or pain. A digital inclinometer was placed at the middle of the posterior thigh halfway between the ishial tuberosity and the popliteal fossa and measured the angle formed from the horizontal. (Figure 3.1)
- **Rectus Femoris**: The participant was positioned prone with her knee flexed to 90 degrees with the hips stabilized to the table using webbing. The contralateral leg was stabilized to the table using webbing. The subject was instructed to relax, and the examiner moved the thigh posteriorly until the point of first resistance or pain. A digital inclinometer was placed at the middle of the posterior thigh halfway between the ishial tuberosity and the popliteal fossa and measured the angle formed from the horizontal. (Figure 3.2)
- **Hip External Rotators**: The participant was positioned in a prone position with his/her knee bent to 90 degrees, so that the shank and foot were perpendicular to the floor, and the femur was in line with the body; the other leg was flat on the table. One researcher stabilized the participant's pelvis by placing a hand on the sacrum then grasp the shank of the leg to be measured with the opposite hand and passively internally rotate the femur until the point of first resistance. Once this point is reached, a second researcher measured the angle, with respect to the

horizontal, with a digital inclinometer placed perpendicular to the length of the lateral fibula (Starkey and Ryan 2002). (Figure 3.3)

- **Hip Internal Rotators**: The participant was positioned in a prone position with his/her knee bent to 90 degrees, so that the shank and foot was perpendicular to the floor, and the femur was in line with the body; the other leg was flat on the table. One researcher stabilized the participant's pelvis by placing a hand on the sacrum, then grasped the shank of the leg to be measured and passively internally rotated the femur until the point of first resistance. Once this point was reached, a second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed perpendicular to the length of the lateral fibula (Starkey and Ryan 2002). (Figure 3.4)
- **Hamstrings at 90-90**: The participant was positioned lying supine with the dominant leg flexed to 90 degrees of hip and knee flexion, with the contralateral leg flat on the table stabilized through the application of webbing. The researcher instructed the participant to actively extend the knee to the point of first compensation. A digital inclinometer was placed onto the mid anterior aspect of tibia and measured the angle formed when compared to the horizontal (Magee 2006). (Figure 3.5)
- **Hip Adductors**: The participant was positioned supine with her legs in full extension flat on the table. One clinician stabilized the contralateral anterior superior iliac spine (ASIS) of the leg being tested and then grasped the medial aspect of the shank and passively abducted the leg until the point of first resistance. This angle was measured with standard goniometer with the stationary

arm over the contralateral ASIS, the fulcrum over the ipsilateral ASIS, and the movement arm over the femur in line with the middle of the patella (Starkey and Ryan 2002). (Figure 3.6)

- **Passive Dorsiflexion**: The participant was positioned supine with the knee straight. The subject was instructed to relax, and the examiner moved the plantar aspect of the foot superiorly until the point of first resistance or pain. The ankle dorsiflexion angle was measured using a goniometer as the angle formed by the shaft of the fibula and the lateral midline of the foot (Piva, Fitzgerald et al. 2006). (Figure 3.7)
- **Weight Bearing Dorsiflexion**: Participants were instructed to place their foot perpendicular to the wall and lunge forward to touch the wall with their knee. The foot was then be moved posteriorly until the maximum range of dorsiflexion is reached, which will be identified by the heel lifting off the ground. A digital inclinometer was placed distal to the tibial tuberosity to measure the angle of the tibia relative to the vertical (Hart, Grindstaff et al. 2009). (Figure 3.8)

Electromyography

Prior to electrode application, each electrode site was identified and marked with a felt tip marker. Each site was shaved using an electric razor and cleaned with a 70% isopropyl alcohol solution to reduce skin impedance. The following muscles and electrode sites were utilized for the study:

- *Gluteus Maximus*: 20% of the distance from the second sacral vertebra to a point 10 cm distal to the greater trochanter, starting from the second sacral vertebra (Ericson, Nisell et al. 1985)

- *Biceps Femoris*: 35% of the distance from the ischial tuberosity to the lateral side of the popliteus cavity, starting from the ischial tuberosity (Rainoldi, Melchiorri et al. 2004)

Each electrode was placed parallel to the orientation of the muscle fibers; one reference electrode was placed over the tibial tuberosity of the ipsilateral tibia. Electrode placement was confirmed with manual muscle testing of each muscle and observation of the muscle activity on an oscilloscope. Once electrode placement was confirmed, the electrodes and leads were secured with omnifix tape. Each respective muscle group (gluteus maximus, biceps femoris) then underwent testing for maximal voluntary isometric contraction (MVIC); three, 5 second isometric holds, with one minute of rest between trials. The MVIC data were used to normalize all EMG activation amplitude data. This was done by dividing the average MVIC activation, averaged over a one second window during the period of greatest EMG activation, by the average EMG activation during the descent phase of the single leg squat. All EMG data were collected at 1000 Hz. The following positions were used for MVIC testing:

- *Gluteus Maximus*: The participant was placed in a prone position with the dominant leg flexed at the knee to 90 degrees and the contralateral leg lying flat on the table. The researcher stabilized the pelvis by placing a hand on the subject's sacrum. The researcher's other hand was placed over the posterior aspect of the participant's thigh, just proximal to the knee joint line. The participant was instructed to maintain the flexed knee position during testing and to attempt to raise his/her thigh off of the testing table while the researcher applied downward pressure (Kendall and McCreary 1993).

- *Biceps Femoris*: The participant was placed in a prone position with the dominant leg flexed at the knee to 90 degrees, the tibia externally rotated, and the nondominant leg lying flat on the table. The researcher stabilized the leg to be tested by placing one hand on the distal 1/3 of the posterior aspect of the thigh. The researcher's other hand grasped the posterior aspect of the dominant leg's heel. The participant was instructed to attempt to pull his/her heel in towards his/her gluteal muscles as the researcher resisted the motion with pressure opposing the motion (Anderson and Hall 2005).

Motion Analysis

A MotionSTAR (Ascension Technologies, Burlington, Vermont) electromagnetic motion capture system was used to collect kinematic data during the squatting and jump-landing tasks. The sensors of the electromagnetic tracking system were placed on the subject's lower extremity at the shank, and thigh, and sacrum. Additionally, a sensor was placed on the participant's spine at the C7 vertebral level. The sensors were secured with athletic pre-wrap and white athletic tape. Following sensor placement, digitization of bony landmarks through a seventh electromagnetic sensor placed on a 5 cm stylus were completed in the following sequence; T12/L1 spineous process, medial femoral epicondyle, lateral femoral epicondyle, medial malleolus, lateral malleolus, left anterior superior iliac spine, right anterior superior iliac spine. Digitization of the bony landmarks established a model template for the participant. Left and right knee joint and ankle centers were calculated as the midpoints between the digitized bony of the medial and lateral femoral condyles and medial and lateral malleoli, respectively. The Bell method was used to approximate the hip joint center (Bell, Brand et al. 1989). Three-dimensional coordinate data were collected at a sampling rate of 140 Hz. One minute of rest will be given between trials.

Overhead Squat

Participants were instructed to stand on a force plate holding her arms over her head and toes facing forward. The participant performed a double leg squat as she descended for one beat of a metronome and then return to the starting position in one beat. The metronome was set at a frequency of 60 beats/minute to control velocity. The participant was instructed as to what constitutes a successful trial, no additional coaching or instructions were given concerning technique. A trial was considered successful if the participant maintained proper form throughout the motion, the task was completed at the correct rate, the heels maintained contact with the ground, and the task was completed in a fluid motion. Subjects performed five practice trials of each task prior to testing. After five practice trials, motion analysis data were collected for five successful squats in succession.

Jump Landing

Participants were positioned on a 30-cm high box placed at a distance ½ of their body height from the leading edge of a force plate with their toes facing forwards. The participant was instructed to land with her dominant foot on the force plate and non-dominant foot off the force plate, and then to immediately jump for maximal vertical height (Bennett, Blackburn et al. 2008). The participant was instructed as to what constitutes a successful trial, no additional coaching or instructions were given concerning technique. A trial was considered successful if the participant maintained proper form throughout the motion and the task was completed in a fluid motion. Subjects performed five practice trials of each task prior to testing. After five practice trials, motion analysis data were collected for five successful trials.

Single Leg Squat

Participants were instructed to stand on a force plate on their dominant leg, with the non-dominant leg flexed at the knee between 90° and 45°, with their hands placed on her hips, with their head, eyes and toes facing forward. The participant was instructed to flex the dominant knee as she descends for one beat of a metronome and then return to the starting position in one beat. The metronome was set at a frequency of 60 beats/minute. The participant was instructed as to what constitutes a successful trial, no additional coaching or instructions were given concerning technique. A trial was considered successful if the participant maintained proper form throughout the motion, the task was completed at the correct rate, the participant did not touch down with the non-dominant foot, the heels maintained contact with the ground, and the task was completed in a fluid motion. Subjects performed five practice trials of each task prior to testing. After five practice trials, motion analysis data was collected for five successful squats in succession.

Isometric Strength Testing:

Participants were strength tested using a handheld dynamometer using standard manual muscle testing procedures commonplace to the sports medicine field. Intersession and intrarater reliability and precision will be established prior to data collection (Table 2).

- **Gluteus Maximus:** The participant was positioned prone with knee flexed to 90 degrees with thigh off table, and the contralateral leg straight. The researcher stabilized the participant's hip by applying pressure on ipsilateral posterior superior iliac spine (PSIS). The participant was instructed to hold leg against applied inferior resistance of the researcher across midbelly of hamstrings halfway between the gluteal fold and the knee joint line, and the peak isometric torque produced was measured with a handheld isometric dynamometer (Hislop and Montgomery 2007). This procedure was repeated with the subject's leg placed into 15 degrees of hip extension verified by digital inclinometer.
- **Gluteus Medius:** The participant was positioned sidelying with the knee straight on the side opposite the leg being tested. The researcher ensured the participant was in alignment with the hip in neutral position with respect to the sagittal plane. The researcher stabilized the pelvis through pressure on the ipsilateral ilium. The subject was then instructed to hold the leg against the applied inferior resistance of the researcher applied directly superior to the lateral joint line of the knee, and the peak isometric torque produced was measured by handheld isometric dynamometer (Hislop and Montgomery

2007). This procedure was repeated with the leg placed in 20 degrees of hip abduction, as verified by digital inclinometer.

DATA PROCESSING AND ANALYSIS

Kinematics and Kinetics:

The Motion Monitor Software (Innovative Sports Training, Inc, Chicago, IL) was used to process the data. A global coordinate system was established where the *x*-axis corresponded with the antero-posterior axis, the *y*-axis corresponded to the medio-lateral axis, and the *z*-axis corresponded to the longitudinal axis. A local coordinate system for each segment was established and aligned with the world axis system after bony landmark digitization. A right handed Euler angle sequence with rotation ordered (Y, X', Z'') was used to calculate joint angles in degrees. Trunk flexion(+)/extension(-) was defined as the trunk relative to vertical about the world *y*-axis. Trunk lateral flexion right (+)/left (-) occured about the world *x*-axis. Trunk rotation left (+)/right (-) occurred about the world *z*-axis. Pelvic tilt: anterior (+)/posterior (-) was defined relative to the trunk segment about the *y*-axis. Hip extension (+)/flexion (-), adduction (+)/abduction (-), internal rotation (+)/external rotation (-) occurred about the pelvis *x, y, z*-axes respectively. Knee flexion (+)/extension (-), varus (+)/valgus (-), and internal (+)/external (-) rotation occurred about the femur's *x, y, z*-axes respectively. The knee joint center position along the world's *y*-axis was used to calculate medial knee displacement during the functional tasks. Inverse dynamics were used to consolidate kinematics with the ground reaction forces to determine internal joint moments. Kinetics were normalized to the product of weight (N), and height (m).

Jump Landing:

Medial knee displacement was calculated as the maximum displacement along the *y*-axis during the loading phase. The loading phase was defined as the time between initial ground contact (vertical ground reaction force >10 N) and peak knee flexion angle. Peak hip extension angle was calculated as the maximum value in the sagittal plane during the loading as well as takeoff stage, which will be defined as time from peak knee flexion until take off (vertical ground reaction force <10 N). Peak trunk forward flexion angle was calculated as the maximum value during the loading stage of the task. Peak anterior pelvic tilt was calculated as the maximum value of anterior superior iliac spine motion during the loading stage of the task.

Single Leg Squat and Overhead Squat

Medial knee displacement was calculated as the maximum knee joint center displacement along the *y*-axis during the descent phase. The descent phase was defined as the time between minimum knee flexion angle and peak knee flexion angle. Peak hip extension angle and moment were calculated as the maximum value throughout the entire task. Peak trunk forward flexion angle was calculated as the maximum value during the descent phase of the task. Peak anterior pelvic tilt was calculated as the maximum value during the descent stage of the task.

Data Reduction

All kinematic data was filtered using a fourth-order low-pass Butterworth filter at 10 Hz. Kinematic data was exported and reduced using a custom computer program. Isometric torque was calculated through force production measured with handheld dynamometry. Torques were normalized to body weight in kilograms.

STATISTICAL ANALYSIS

Independent samples t-tests were performed to determine the differences in lower extremity range of motion, gluteus maximus and medius strength, and lower extremity kinetics and kinematics during functional tasks between the “restricted” and “normal” hip flexor groups. Statistical significance will be set at $\alpha<0.05$. All data will be analyzed using SPSS 19.0 statistical software (SPSS, Inc., Chicago, IL).

CHAPTER IV

MANSCRIPT

INTRODUCTION

Rupture of the Anterior Cruciate Ligament (ACL) is a common, costly and debilitating injury. It is estimated that there are approximately 100,000 to 250,000 ACL injuries per year (Toth and Cordasco 2001; Marshall, Padua et al. 2007), accounting for approximately 1 million physician visits annually in the United States (Marshall, Padua et al. 2007). The estimated financial burden is conservatively estimated at $11,500 per patient, which leads to approximately $1.5 billion dollars in medical costs per year (Gottlob C. A. 2000). Thirty seven percent of patients never return to the same level of activity (Ardern, Webster et al.), with only 44% returning to competitive sports, and 18% never returning to any sport activity (Ardern, Webster et al.). The average time from surgery to return to activity is 7.3 months (Ardern, Webster et al.). Furthermore, there is a serious psychological toll placed on those with ACL injury, as those who sustain ACL injury have a higher depression rate for a longer duration than those who suffer concussion (Mainwaring, Hutchison et al. 2010). Given the frequency of ACL injury and it's many negative consequences it is important to understand factors that may predispose individuals to future ACL injury

Several factors are theorized to predispose individuals to ACL pathology; however, lumbo-pelvic hip complex and frontal plane knee biomechanics are both significantly related to ACL injury and mechanism (Zazulak, Ponce et al. 2005; Hewett, Myer et al. 2006). Ireland identified the “position of no return” in which the hip abductors and extensors “shut down”, and fail to effectively control lumbo-pelvic-hip motion. This results in hip internal rotation, hip adduction, knee valgus, and tibial external rotation (Ireland 1999). Zazulak also noted an association between proprioception of the trunk and neuromuscular control during ACL injury incidents. Furthermore, it was noted that uncontrolled or excessive trunk motion was found in ACL injury events(Alentorn-Geli, Myer et al. 2009; Hewett, Torg et al. 2009; Kulas, Hortobagyi et al. 2010; Hewett T 2011). Knee valgus, and by association, medial knee displacement are frequently described as a biomechanical risk factor for ACL pathology (Hruska 1998; Delp, Hess et al. 1999; Boden, Griffin et al. 2000; Hertel J 2004; Agel, Arendt et al. 2005; Alentorn-Geli, Myer et al. 2009; Chiaia, Maschi et al. 2009).

Hip musculature weakness and/or muscle activation is linked to greater anterior pelvic tilt (Popovich and Kulig). Greater anterior pelvic tilt has, in turn, been correlated with ACL injury (Hertel J 2004), and has been theorized to be caused by overactive/tight hip flexors. As such, this anterior pelvic tilting has also been associated with increased hip internal rotation, likely due to alterations of the moment arm of the deep hip external rotators (Delp, Hess et al. 1999), which may force the body into Ireland’s position of no return. Another theory revolves around an inhibition of the gluteus maximus due to either an altered length tension relationship, or through a reciprocal inhibition. This would then limit the force production capability of the gluteus maximus, and then potentially lead to altered biomechanics during activity.

Previous studies have examined other aspects of hip and trunk biomechanics, and have linked it to ACL pathology risk factors, including lateral trunk motion, overactive hip adductors, and underactive or inhibited hip abductors (Hewett T 2011). Furthermore, greater flexibility of the hamstrings and lesser activation has been suggested to decrease knee stiffness and increase knee valgus (Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003).

Furthermore, hip flexor restriction has been theorized to affect lumbo-pelvic hip motion, as well as ACL injury risk factors, including medial knee displacement, excessive trunk forward flexion, and femoral internal rotation. Female soccer players were observed to possess decreased hip flexor flexibility as measured by a “Thomas test” compared to normative values for a similar demographic that does not play soccer (Chiaia, Maschi et al. 2009). Delp et al identified that greater hip flexion is associated with a shifting of the line of pull of the external rotators, which, in turn, forces these muscle to cause an internal rotation moment of the hip (Delp, Hess et al. 1999). Hollis et. al (1991) first noted that knee valgus was linked to hip internal rotation as weight bearing knee flexion angles increase during a squatting task (Hollis, Takai et al. 1991). Shin et. al (2011) also observed that knee valgus in association with hip internal rotation increased the strain on the ACL more than either motion individually (Shin, Chaudhari et al. 2011). Femoral internal rotation has also been observed to be a risk factor for ACL pathology when combined with other rotational movements (Shimokochi and Shultz 2008; Alentorn-Geli, Myer et al. 2009). This was compounded by the finding that female soccer players were observed to have greater femoral internal rotation compared to normative values of similar demographics (Chiaia, Maschi et al. 2009), as well as when compared to males of similar demographics (Brophy, Chiaia et al. 2009). As such, further examination of the effects of hip flexor tightness and its relationship to lower extremity biomechanics and ACL injury is indicated.

Therefore, the purpose of this study is to examine the difference in biomechanical risk factors for ACL injury during three dynamic tasks between female soccer players with restricted hip flexors and those with "normal" hip flexor length. Determining this relationship will allow for clinicians to screen for a potential risk factor for ACL pathology, and could potentially lead to prevention strategies through correction of muscle imbalances that characterize hip flexor muscle restriction.

METHODS

SUBJECTS

A total of 40 females were selected from the lifetime fitness classes, women's intramural, club, and varsity soccer teams, at the University of North Carolina at Chapel Hill. The tight hip flexor group consisted of 20 females, while the control group consisted of 20 females with no significant difference between groups for height, weight, or age. Group sizes were based on power calculations for an estimated power calculation of 0.80 based on effect size from previous studies (Ford, Myer et al. 2003; McLean, Huang et al. 2005; Pantano, White et al. 2005). Subjects were required to play soccer at least twice a week for one hour at a time to qualify. Subjects with a history of lower extremity surgery or fracture were excluded.

The participants read and signed an informed consent form approved by the Institutional Review Board (IRB) of the University of North Carolina at Chapel Hill. All participants also completed a questionnaire to confirm inclusion/exclusion criteria, the participant's dominant leg (the leg that they would use to kick a soccer ball for maximum distance), medical history, and contact information.

Participants underwent a screening process to determine group assignment through completing a modified Thomas Test, which was performed with the subject lying supine on a table with both legs held tight to chest. The examiner then slowly lowered the participant's leg to the point of first resistance, while stabilizing the pelvis. A digital inclinometer was placed along anterior aspect of thigh between ASIS and patella halfway from the superior pole of patella to ASIS in order to determine hip flexor tightness. Subjects were classified based on normative values of greater than or equal to 0° above the horizontal axis for "restricted", and greater than or equal to 15° below horizontal for "control"(Ferber, Kendall et al. 2010) across the average of 2 trials. The modified Thomas test has been shown to have good reliability (Gabbe, Bennell et al. 2004; Peeler and Anderson 2007; Clapis, Davis et al. 2008). Participants' height and weight were then recorded. Participants then completed a warm-up on a stationary cycle ergometer at a self-selected pace for five minutes at a rate of perceived exertion of 3/10.

INSTRUMENTATION

An electromagnetic motion tracking system sampling at 140 Hz (trakSTAR, Ascension Technologies, Inc, Burlington, VT) and non-conductive force plate sampling at 1400 Hz (Bertec 4060-NC, Columbus, OH) were used to collect lower extremity kinematics and ground reaction force data, respectively. The motion tracking systemdevice consists of an extended range transmitter that emits an electromagnetic field and standard receivers (dimensions 25.4 X 25.4 X 20.3 mm) that detect the electromagnetic field.

Joint angles for measures of flexibility of the rectus femoris, iliopsoas, hip internal rotators, hip external rotators, ankle plantarflexors, and hamstrings were measured using a digital inclinometer (Saunders Group, Inc., Chaska, MN). Intersession and intrarater reliability of the active range of motion testing procedure of the investigator responsible for taking the measures in this study were calculated with intraclass coefficients (ICC) and standard errors of the measurement (SEM) for each range of motion measurement (ICC 3,k range 0.996-0.965, SEM Range 0.2502-2.024) (Table 1). A digital inclinometer has been found to be reliable and valid for these measurements (Bierma-Zeinstra, Bohnen et al. 1998).

Joint angles for measures of flexibility of the hip adductors and plantar flexors were measured using a standard 30.5 cm (12 in) plastic goniometer. Intersession and intrarater reliability of the passive range of motion testing procedure of the investigator responsible for taking the measures in this study was calculated with intraclass coefficients (ICC) and standard error of the measurement (SEM) for each range of motion measurement (ICC 3,k range, .909-.992; SEM range, 0.836-1.099) (Table 1).

A handheld digital dynamometer (Chatillon MSC-500, AMETEK, Inc., Largo, FL) was used to collect peak and mean isometric strength values for the gluteus maximus and gluteus medius. It has been found to be reliable for measurement of hip extension strength (ICC 0.75-0.85) (van der Linden, Aitchison et al. 2004), as well as valid for these measurements (Reed, Den Hartog et al. 1993; Trudelle-Jackson, Jackson et al. 1994). Intersession and intrarater reliability of the isometric strength testing were calculated using intraclass coefficients (ICC) and standard error of the measurement (SEM) for each muscle tested (ICC 3,k range 0.862-0.979, SEM Range 0.477-1.445) (Table 2).

Surface electromyography (EMG) data (Bagnoli-8; Delsys, Inc, Boston, MA) were collected with an amplification factor of 1,000 (20 – 45 Hz), common-mode rejection ratio of 60 Hz (>80 dB), and input impedance > 1015//0.2 Ω//pF was used to record lower extremity muscle activity.

PROCEDURES

RANGE OF MOTION MEASUREMENTS

All range of motion measurements were measured with a digital inclinometer or standard goniometer. Intersession and intrarater reliability and precision were established prior to data collection (Table 1). For each of the following muscle groups, the examiner moved the participant until point of first resistance or compensation from accessory motion was noted. Two trials were taken for each range of motion measurement using standard range of motion techniques(Starkey and Ryan 2002). Measurements were taken of rectus femoris, iliopsoas, plantarflexors (passive and lunge) (Hart, Grindstaff et al. 2009), hip adductors, hip internal and external rotators, and hamstrings at 90/90(Magee 2006).

ELECTROMYOGRAPHY

Prior to electrode application, the electrode sites for the biceps femoris and gluteus maximus were identified (Ericson, Nisell et al. 1985; Rainoldi, Melchiorri et al. 2004). Each site was shaved using an electric razor and cleaned with a 70% isopropyl alcohol solution to reduce skin impedance. A reference electrode was placed at the tibial tuberosity of the dominant leg. Surface electrodes were placed over the biceps femoris and gluteus maximus muscles with an interelectrode distance of 10 mm. The electrode for the biceps femoris was placed between the ishial tuberosity and lateral popliteal cavity approximately 1/3 of the way distal from the ishial tuberosity(Rainoldi, Melchiorri et al. 2004). The electrode for the gluteus maximus was placed between the second sacral vertebrae and the greater trochanter approximately 1/3 of the way from the sacrum (Rainoldi, Melchiorri et al. 2004).

Each respective muscle group then underwent testing for maximal voluntary isometric contraction (MVIC); three, 5 second isometric holds, with one minute of rest between trials. All EMG data were collected at 1000 Hz.

MOTION ANALYSIS

Kinematic and kinetic data were collected during the squat and jumping tasks. The sensors of the electromagnetic tracking system were placed on the subject's lower extremities at the shank, and thigh, and sacrum, as well as the C7 spineous process. Following sensor placement, digitization of bony landmarks was performed. Left and right knee joint and ankle centers were calculated as the midpoints between the digitized bony of the medial and lateral femoral condyles and medial and lateral malleoli, respectively. The Bell method was used to approximate the hip joint center (Bell, Brand et al. 1989).

SQUATTING TRIALS

Participants were instructed to stand on a force plate holding arms over her head and toes facing forward. The participant performed a double leg squat as she descended for one beat of a metronome set at 60 beats per minute and then returned to the starting position in one beat. The participant was instructed as to what constituted a successful trial, no additional coaching or instructions were given concerning technique. A trial was considered successful if the participant maintained proper form throughout the motion, the task was completed at the correct rate, the heels maintained contact with the ground, and the task was completed in a fluid motion. Subjects performed five practice trials, then data was collected for five consecutive squats.

ISOMETRIC STRENGTH TESTING:

Muscle testing was performed according to standard manual muscle testing procedures (Hislop and Montgomery 2007). Peak force was measured with a handheld dynamometer.

DATA ANALYSIS

The Motion Monitor Software (Innovative Sports Training, Inc, Chicago, IL) was used to collect the kinematic and surface electromyography data. A right-handed Euler angle sequence with rotation ordered (Y, X', Z'') was used to calculate joint angles in degrees with the motion defined as the distal segment in relation to the proximal segment. Triplanar segment angles of the trunk and pelvis relative to the world reference frame were used to describe trunk and pelvis kinematics. The knee joint center position along the world's *y*-axis was used to calculate medial knee displacement during the functional tasks. All variables of interest were assessed during the descent phase of the overhead squat task; defined as the time between minimum knee flexion angle and peak knee flexion angle during the squat.

Peak triplanar trunk, pelvis, hip, and knee kinematics were calculated during the descent phase of the overhead squat. Joint and trunk and pelvis segment angular displacements were calculated as the difference between the maximal angular position of the joints and segments and the position of the joint at the start of the descent phase of the squat task. Medial knee displacement was calculated as the difference between the maximum medial position of the knee joint center along the *y*-axis during the descent phase of the squatting task and the position of the knee joint center along the *y*-axis at the start of the descent phase. Static pelvic tilt position was calculated as the average angular position of the pelvis rigid body segment at the point 100 milliseconds prior to initiation of the initial squat during the overhead squat trials.

All kinematic data were filtered using a fourth-order low-pass Butterworth filter at 10 Hz. Kinematic data was exported and reduced using a custom computer program using Matlab version R2012b (Mathworks Inc, Natick, MA). Isometric torque was calculated through force production measured with handheld dynamometry (Lafayette Manual Muscle Tester, Model 01163, Lafayette, IN). Torques were then be normalized to body weight in kilograms.

STATISTICAL ANALYSIS

Independent samples t-tests were performed to determine the differences in lower extremity range of motion, gluteus maximus and medius strength, and lower extremity kinetics and kinematics during functional tasks between the "restricted" and "normal" hip flexor groups. Statistical significance was set at $\alpha<0.05$. All data were analyzed using SPSS 19.0 statistical software (SPSS, Inc., Chicago, IL).

RESULTS

There was no difference between restricted and control participants for height ($p=0.830$) and mass ($p=0.150$). The inclusion criteria resulted in groups that were significantly different in their modified Thomas Test values of hip extension ($p<0.001$, $d=2.18$). Thus, the inclusion criterion was successful in capturing subjects with restricted hip flexor length.

RANGE OF MOTION

In addition to decreased hip extension during the modified Thomas Test, several range of motion measures were also significantly decreased in the restricted group compared to the control group. Subjects with restricted hip flexors had significantly less ankle dorsiflexion as measured by the lunge test (mean difference 5.628°, $p=0.041$, $d=-0.711$), and passive dorsiflexion with the knee extended (mean difference 9.625°, $p<0.001$, $d=-1.32$). Restricted subjects also demonstrated significantly less hip abduction (mean difference 6.200° $p=0.007$, $d=-0.984$), hip external rotation (mean difference

6.338°, p=0.027, d=-0.800), hip extension with the knee straight (mean difference 10.115°, p<0.001, d=-1.53), and hip extension with the knee flexed (mean difference 10.538°, p<0.001, d=-1.40). There was no significant difference in hamstring flexibility measured at 90/90 (mean difference 5.813° p=0.055, d=0.579), or hip internal rotation (mean difference 1.660°, p=0.592, d=-0.172). Means and standard deviations are noted in Table 3.

STRENGTH

GLUTEUS MAXIMUS

There was no significant difference in gluteus maximus strength at neutral (mean difference 0.01, p=0.73, d=0.09). There was also no significant difference in gluteus medius strength at neutral (mean difference 0.04, p=0.56 d=0.275) Means and standard deviations are noted in Table 4.

ELECTROMYOGRAPHY

Subjects with restricted hip flexors displayed significantly less average gluteus maximus activation during the overhead squatting task compared to control subjects (mean difference 0.06, p=0.020, d=-0.783). There was no significant difference in average biceps femoris activation between groups (mean difference 0.23, p=0.35, d=0.306). Means and standard deviations are noted in Table 5.

KINEMATICS

Subjects with restricted hip flexors were observed to have greater posterior pelvic tilt in a static posture (mean difference 5.05, p=0.014, d=0.881). Mean and standard deviation are noted in Table 6.

There were no differences in peak hip flexion angle (mean difference 1.01, p=0.857, d=0.056), peak knee flexion angle (mean difference 2.38, p=0.684, d=0.130), trunk flexion (mean difference 5.77, p=0.877, d=0.391), or anterior pelvic tilt (mean difference 1.81, p=0.552, d=0.084). There were also no differences in peak hip adduction (mean difference 3.34, p=0.261, d=0.361), peak knee valgus (mean difference 1.26, p=0.739, d=0.185). Means and standard deviations are noted in Table 6.

Means and Standard Deviations for joint displacement during the overhead squat are listed in Table 7. During the overhead squatting task, subjects with restricted hip flexors were observed to have less hip adduction displacement (mean difference 2.70°, p=0.022, d=0.748), less knee internal rotation displacement (mean difference 5.54°, p=0.009, d=0.876), and less trunk rotation displacement towards the dominant limb (mean difference 1.86°, p=0.006, d=-0.917). However, there were no significant differences noted in trunk flexion displacement (mean difference 3.06, p=0.307, d=-0.375), anterior pelvic tilt displacement (mean difference 0.81, p=0.391, d=0.096), hip flexion displacement (mean difference 1.44, p=0.783, d=0.087), or knee flexion displacement (mean difference 0.71, p=0.895, d=0.042). There were also no differences noted in knee valgus displacement (mean difference 4.45, p=0.521, d=-0.650) or medial knee displacement (mean difference 0.014m, p=0.111, d=0.509).

DISCUSSION

The most important finding was that the restricted hip flexor group demonstrated less gluteus maximus muscle activation during the overhead squat compared to the control group. In addition, the restricted hip flexor group displayed less ankle dorsiflexion, hip abduction, and hip external rotation flexibility compared to those who have "normal" hip flexor length as measured by a Modified Thomas Test. Thus, individuals with restricted hip flexors have altered neuromuscular control of the gluteus maxiums combined with range of motion restrictions throughout the posterior kinetic chain (e.g. hip rotators, hip adductors, hamstrings, gastrocnemius/soleus). Differences in gluteus maximus activation and posterior kinetic chain range of motion restrictions may have contributed to our observations of significantly less trunk rotation, hip adduction and knee internal rotation displacement in the restricted hip flexor group compared to the control group during the overhead squat task.

The lack of gluteus maximus activation during an overhead squat in the restricted group has large injury implication, as gluteal activation has been noted repeatedly with respect to poor biomechanical movements during functional tasks (Popovich and Kulig ; Hollman, Ginos et al. 2009; Homan 2011; Hollman, Hohl et al. 2012). This is also important, as gluteal control is also responsible for neuromuscular control, which is also associated with lower extremity injury (Boden, Griffin et al. 2000; Griffin, Agel et al. 2000; Lephart, Abt et al. 2002; Hewett, Myer et al. 2006; Kernozek, Torry et al. 2008).

The decreased gluteus maximus activation noted in our study may be due to the concept of reciprocal inhibition (Moore and Hutton 1980; Alter 1996; Liebenson 1996).

This theory states that muscles that are “tight” or “restricted” are also overactive. This over-activity of the agonist muscle group then leads to an inhibition of the antagonist group. This causes the body to compensate to perform a physiologic action through the use of accessory muscles.

Under this proposed theory, restricted (or overactive) hip flexors would cause a neural inhibition of the gluteal muscles. This would cause an increased reliance on the synergistic hip extensor muscles, such as the hamstrings and adductors to assist with the physiologic motion of hip extension. Our findings of decreased gluteus maximus activation in the restricted group (0.081 ± 0.047) compared to the control group (0.141 ± 0.098) indicate that there appears to be an inhibition of the gluteus maximus in subjects with restricted hip flexors. This is combined with difference in lateral hamstring activation in the restricted group (0.13 ± 0.12) compared to the control group (0.071 ± 0.04) suggests greater hamstrings activation between groups, though there was no statistical significance.

Inspection of these values demonstrates that the restricted group appeared to have greater relative activation of the hamstrings compared to the gluteus maximus. In contrast, the control group appeared to have over 2 times greater gluteus maximums relative to hamstrings activation. This is further evident as subjects in the restricted group had greater biceps femoris activity compared to gluteal activity during an overhead squat (0.13 vs. 0.081), compared to control subjects, who displayed greater gluteus maximus activity compared to biceps femoris activity (0.141 vs. 0.071). Thus, the restricted group appears to use a synergistic dominance muscle activation strategy with greater reliance on the hamstrings and less on the gluteus maximums.

In combination with the apparent inhibition of the gluteus maximus and synergistic dominance of the hamstrings, there appears to be additional biomechanical changes in the posterior lower extremity. These findings include restriction of a wide array of posterior muscles, including the internal rotators, a trend towards hamstring tightness, as well as restriction of the gastrocnemius and soleus complex. Restricted subjects also displayed restricted hip adductors compared to the control subjects. These findings are significant as a lack of dorsiflexion has been implicated in ACL injury risk (Rabin and Kozol 2010; Fong, Blackburn et al. 2011), and hip adduction and internal rotation are both cited as part of Ireland's "position of no return"(Ireland 1999). As such, the findings could possibly be indicative of an increased injury risk for those subjects with a positive Thomas Test.

The difference in anterior pelvic tilt may be due to the posterior tightness that was observed along with the hip flexor restriction throughout the lower extremity, as the internal rotators, biceps femoris, and gastrocnemius were all found to be restricted in those with hip flexor restriction. Through the myofascial connections in the posterior lower extremity, tightness posteriorly may lead to a posterior pull on the pelvis, which, in turn, would limit the rotation of the pelvis, and thus, limit the potentially harmful effects down the kinetic chain mechanically. This is supported by Lopez-Minarro et al, who found that through stretching the hamstrings, there was a decrease in anterior pelvic tilt compared to a control group (Lopez-Minarro, Muyor et al. 2012). While gluteus maximus activation was decreased in the restricted group, the line of pull from the hamstrings provides a greater mechanical advantage (Neuman 2010), which thus may explain the difference in posterior pull between groups.

Furthermore, our findings of internal rotator tightness, combined with lateral hamstring and lateral gastrocnemius tightness have been theorized to facilitate a varus movement dysfunction in the lower extremity. Particularly due to the lateral attachments and the myofascial connections along the lateral leg, these muscles in combination would likely pull the pelvis in a posterior direction, as well as facilitate hip and knee external rotation, which, in turn are supported by our results, as we noted decreased knee internal rotation displacement in the control group during the overhead squatting task.

These findings are contrary to our initial hypothesis, as our hypothesis was based primarily on a biomechanical model based exclusively on hip flexor tightness. However, upon examination, the combination of hip flexor tightness and lower extremity posterior restriction provides very different movement patterns, particularly macroscopically upon examination by a clinician. This may be due to a reciprocal inhibition, or though a potential synergistic dominance, which may be to accomplish a physiologic motion, or may be protective in nature to attempt and limit rotation of the pelvis in the sagittal plane.

There were no differences in isometric strength noted for any muscles tested between groups. As such, it appears that individuals with hip flexor restriction do not have strength deficits, but instead have altered gluteus maximus muscle activation combined with posterior chain tightness, which appears to influence movement patterns during an overhead squat.

The differences between the restricted hip flexor and control groups in kinematics during the functional tasks seems to indicate that further study is required in order to determine the true clinical significance of the modified Thomas Test with respect to ACL injury, to determine any potential protective mechanisms that may occur, as well as the examination of male athletes and across more diverse populations. Furthermore, our study did not examine subjects with isolated hip flexor restriction, as such; subjects with those restrictions may display very different movement patterns. As such, our results speak to the need for follow up evaluation, as well as assessment for global lower extremity tightness when examining a patient in the clinic.

FIGURES

FIGURE 1.1: SELECTION OF CONTROL GROUP

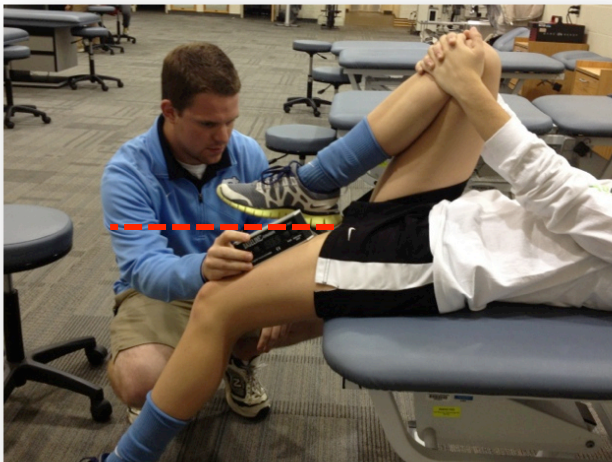

FIGURE 1.2: SELECTION OF RESTRICTED GROUP

FIGURE 3.1: MEASURE OF ILIOPSOAS TIGHTNESS

FIGURE 3.2: MEASURE OF RECTUS FEMORIS TIGHTNESS

FIGURE 3.3: MEASURE OF HIP EXTERNAL ROTATION

FIGURE 3.4: MEASURE OF HIP INTERNAL ROTATION

FIGURE 3.5: MEASURE OF HAMSTRINGS AT 90/90

FIGURE 3.6: MEASURE OF HIP ABDUCTION

FIGURE 3.7: MEASURE OF PASSIVE DORSIFLEXION

FIGURE 3.8: MEASURE OF WEIGHT BEARING DORSIFLEXION

TABLES

TABLE 1: INTRACLASS CORRELATION COEFFICIENTS AND STANDARD ERROR OF THE MEASUREMENT FOR PASSIVE RANGE OF MOTION MEASUREMENTS

Measure	ICC	SEM
Thomas Test	0.9935	0.8485
Passive Dorsiflexion	0.9929	0.8454
Lunge Test	0.9945	0.2502
Hamstrings at 90/90	0.9895	1.5550
Hip Internal Rotation	0.9856	0.7070
Hip External Rotation	0.9962	1.2245
Hip Abduction	0.9859	1.0991
Hip Extension Knee Flexed	0.9652	1.1083
Hip Extension Knee Straight	0.9940	0.8648

TABLE 2: INTRACLASS CORRELATION COEFFICIENTS AND STANDARD ERROR OF THE MEASUREMENT FOR ISOMETRIC STRENGTH MEASUREMENTS

Measure	ICC	SEM
Gluteus Maximus	0.9753	0.4767
Gluteus Medius	0.9624	1.4449

TABLE 3: MEANS AND STANDARD DEVIATIONS FOR RANGE OF MOTION VARIABLES

Measure	Restricted ± SD	Control ± SD	p-Value
Thomas Test	12.85° ± 5.05°	-19.52° ± 9.19°	<0.001**
Lunge Test	45.13° ± 5.80°	50.75° ± 10.23°	0.041
Passive Dorsiflexion	3.08° ± 5.99°	12.70° ± 5.54°	<0.001**
Hamstring at 90/90	-16.96° ± 10.52°	-11.14° ± 7.88°	0.055
Abduction	49.75° ± 6.71°	55.95° ± 6.94°	0.007**
Hip External Rotation	41.18° ± 6.88°	47.52° ± 10.20°	0.027*
Hip Internal Rotation	35.50° ± 8.72°	37.16° ± 10.60°	0.592
Hip Extension Knee Straight	21.90° ± 4.97°	32.44° ± 6.32°	<0.001**
Hip Extension Knee Flexed	25.94° ± 5.96°	36.05° ± 5.67°	<0.001**

*=Significant at p<0.05 level
**- Significant at p<0.01 level

TABLE 4: MEANS AND STANDARD DEVIATIONS FOR STRENGTH VARIABLES

Measure	Restricted ± SD	Control ± SD	p-Value
Gluteus Max. at Neutral	0.39 ± 0.21	0.40 ± 0.14	0.73
Gluteus Med. at Neutral	0.50 ± 0.28	0.46 ± 0.17	0.56

TABLE 5: MEANS AND STANDARD DEVIATIONS FOR EMG DATA OF HIP EXTENSORS DURING AN OVERHEAD SQUATTING TASK

Measure	Restricted ± SD	Control ± SD	p-Value
GMAX	0.081 ± 0.047	0.141 ± 0.098	0.020*
Biceps Fem.	0.13 ± 0.12	0.071 ± 0.04	0.058

*=Significant at p<0.05 level
**- Significant at p<0.01 level

TABLE 6: MEANS AND STANDARD DEVIATIONS FOR PEAK KINEMATIC VARIABLES DURING AN OVERHHEAD SQUATTING TASK

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-81.71° ± 16.24°	-82.72° ± 19.18°	0.857
Hip Frontal Plane	-15.72° ± 8.94°	-12.38° ± 9.55°	0.261
Hip Transverse Plane	-13.94° ± 9.55°	-13.40° ± 13.04°	0.883
Knee Sagittal Plane	94.96° ± 14.68°	92.58° ± 21.40°	0.684
Knee Frontal Plane	-6.99° ± 5.55°	-8.25° ± 7.89°	0.563
Knee Transverse Plane	-6.66° ± 10.49°	-7.61° ± 6.85°	0.739
Trunk Sagittal Plane	12.16° ± 14.48°	17.93° ± 15.02°	0.877
Trunk Frontal Plane	-5.76° ± 8.25°	-9.41° ± 6.07°	0.119
Trunk Transverse Plane	-0.86° ± 9.21°	-2.69° ± 12.68°	0.604
Pelvis Sagittal Plane	27.30° ± 29.11°	29.11° ± 8.85°	0.552
Pelvis Frontal Plane	-3.68° ± 5.70°	-3.37° ± 7.42°	0.881
Static Anterior Pelvic Tilt	-0.12° ± 5.60°	4.93° ± 5.86°	0.014*

*=Significant at $p<0.05$ level
**- Significant at $p<0.01$ level

TABLE 7: MEANS AND STANDARD DEVIATIONS FOR DISPLACMENT KINEMATIC VARIABLES DURING AN OVERHEAD SQUATTING TASK

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-77.53° ± 14.78°	-78.97° ± 17.88°	0.783
Hip Frontal Plane	1.85° ± 2.30°	4.55° ± 4.14°	0.022*
Hip Transverse Plane	-5.38° ± 4.31°	-7.91° ± 5.03°	0.097
Knee Sagittal Plane	96.46° ± 15.56°	95.75° ± 18.07°	0.895
Knee Frontal Plane	-9.20° ± 8.15°	-4.75° ± 5.20°	0.521
Knee Transverse Plane	7.41° ± 6.17°	12.94° ± 6.45°	0.009**
Medial Knee Displacement	0.056m ± 0.198m	0.070m ± 0.334m	0.089
Trunk Sagittal Plane	21.95° ± 9.26°	25.01° ± 6.87°	0.307
Trunk Frontal Plane	-1.42° ± 1.01°	-1.23° ± 1.25°	0.377
Trunk Transverse Plane	-2.21° ± 1.31°	-4.07° ± 2.55°	0.006**
Pelvis Sagittal Plane	20.13° ± 9.04°	20.94° ± 7.65°	0.391
Pelvis Frontal Plane	1.97° ± 1.54°	1.44° ± 1.47°	0.630

*=Significant at $p<0.05$ level
**- Significant at $p<0.01$ level

APPENDIX ONE: ADDITIONAL RESULTS

Jump Landing Kinematics and Kinetics

During the jump-landing task, subjects with restricted hip flexors were observed to land in a more adducted position at the knee (mean difference 3.52°, $p=0.041$), and had more displacement into trunk flexion across the loading phase (mean difference 14.68°, $p<0.0005$). There were no differences noted in peak frontal or sagittal plane kinematics. Means and Standard Deviations are noted in Tables A.1-A.3. Subjects with restricted hip flexors were also observed to have a lower hip abduction moment compared to those in the control group (mean difference 0.025°, $p=0.041$). There were no differences in hip sagittal kinetics, or knee kinetics between groups. Means and standard deviations are noted in Table A.4.

Table A.1: Means and Standard Deviations for IGC Kinematic Variables During a Jump Landing Task

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-34.11° ± 12.31°	-32.05° ± 10.85°	0.578
Hip Frontal Plane	-8.38° ± 1.93°	-8.47° ± 9.25°	0.966
Knee Sagittal Plane	23.08° ± 11.01°	19.23° ± 10.05°	0.255
Knee Frontal Plane	-0.94° ± 6.27°	2.58° ± 4.01°	0.041*
Trunk Sagittal Plane	18.30° ± 11.20°	18.93° ± 10.84°	0.858
Trunk Frontal Plane	-3.25° ± 9.76°	-6.86° ± 6.63°	0.180
Pelvis Sagittal Plane	5.73° ± 4.18°	6.76° ± 6.98°	0.579
Pelvis Frontal Plane	-2.30° ± 6.75°	-2.64° ± 6.48°	0.869

*=Significant at p<0.05 level

Table A.2: Means and Standard Deviations for Peak Kinematic Variables During a Jump Landing Task

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-75.9° ± 19.31°	-71.70° ± 21.20°	0.522
Hip Frontal Plane	-12.84° ± 9.20°	-10.53° ± 8.72°	0.420
Knee Sagittal Plane	97.60° ± 17.78°	88.20° ± 21.67°	0.142
Knee Frontal Plane	-7.13° ± 10.24°	-9.59° ± 9.31°	0.431
Trunk Sagittal Plane	32.84° ± 13.84°	27.72° ± 11.81°	0.220
Trunk Frontal Plane	-3.13° ± 9.32°	-5.38° ± 8.26°	0.136
Pelvis Sagittal Plane	15.81° ± 8.80°	17.88° ± 12.73°	0.552
Pelvis Frontal Plane	-5.17° ± 8.94°	-8.81° ± 5.88°	0.859

Table A.3: Means and Standard Deviations for Displacement Kinematic Variables During a Jump Landing Task

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-41.99° ± 16.62°	-39.33° ± 18.29°	0.614
Hip Frontal Plane	-5.78° ± 4.64°	-4.67° ± 3.02°	0.377
Knee Sagittal Plane	75.06° ± 12.50°	68.79° ± 16.16°	0.178
Knee Frontal Plane	-10.28° ± 7.27°	-10.86° ± 7.27°	0.812
Medial Knee Displacement	0.04m ± 0.01m	0.04m ± 0.01m	0.810
Trunk Sagittal Plane	10.95° ± 10.32°	15.31° ± 10.49°	0.193
Trunk Frontal Plane	-4.69° ± 4.14°	-4.06° ± 2.50°	0.603

Table A.4: Means and Standard Deviations for Peak Kinetic Variables During a Jump Landing Task

Measure	Restricted Mean ± SD	Control Mean ± SD	p-Value
Hip Extension Moment	-0.24 ± 0.085	-0.22 ± 0.10	0.431
Hip Abduction Moment	-0.015 ± 0.027	-0.040 ± 0.047	0.041*
Knee Extension Moment	-0.20 ± 0.037	-0.19 ± 0.064	0.420
Knee Varus Moment	-0.013 ± 0.025	-0.016 ± 0.026	0.741

*=Significant at p<0.05 level
**- Significant at p<0.01 level

Single Leg Squat Kinematics and Kinetics

During the single leg squatting task, subjects with restricted hip flexors were observed to have lower peak hip frontal plane angle (mean difference 7.89°, p=0.017), and had less hip displacement in the frontal plane (mean difference 8.54°, p=0.002). There were no differences noted in sagittal plane kinematics. There were also no differences in knee, pelvis, or trunk kinematics. Means and Standard Deviations are noted in Tables A.5-A.6. Subjects with restricted hip flexors were also observed to have a hip abduction moment compared to a hip adduction moment in the control group (mean difference 0.051, p<0.0005). There were no differences in hip sagittal kinetics, or knee kinetics between groups. Means and standard deviations are noted in Table A.7.

Table A.5: Means and Standard Deviations for Peak Kinematic Variables During a Single Leg Squat

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-49.84° ± 19.56°	-53.83° ± 17.69°	0.503
Hip Frontal Plane	14.69° ± 10.55°	22.58° ± 9.36°	0.017*
Knee Sagittal Plane	66.29° ± 9.57°	65.61° ± 15.08°	0.866
Knee Frontal Plane	-5.60° ± 5.30°	-5.44° ± 7.31°	0.936
Trunk Sagittal Plane	16.79° ± 13.08°	21.48° ± 12.11°	0.247
Trunk Frontal Plane	-4.21° ± 7.15°	-7.47° ± 6.35°	0.136
Pelvis Sagittal Plane	20.90° ± 12.86°	23.72° ± 10.48°	0.451
Pelvis Frontal Plane	-3.89° ± 7.31°	-6.87° ± 9.71°	0.279

*=Significant at p<0.05 level

Table A.6: Means and Standard Deviations for Displacement Kinematic Variables During a Single Leg Squat

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Sagittal Plane	-44.65° ± 17.65°	-50.05° ± 14.29°	0.295
Hip Frontal Plane	13.49° ± 7.94°	22.03° ± 8.21°	0.002**
Knee Sagittal Plane	64.54° ± 10.84°	66.27° ± 10.21°	0.608
Knee Frontal Plane	-8.85° ± 7.37°	-9.83° ± 8.78°	0.704
Medial Knee Displacement	0.04m ± 0.01m	0.04m ± 0.01m	0.129
Trunk Sagittal Plane	12.02° ± 7.76°	16.24° ± 6.60°	0.072
Trunk Frontal Plane	2.54° ± 2.61°	3.39° ± 3.06°	0.346
Pelvis Sagittal Plane	12.81° ± 10.67°	15.81° ± 8.95°	0.341
Pelvis Frontal Plane	0.80° ± 0.66°	0.56° ± 0.53°	0.226

*=Significant at p<0.05 level
**- Significant at p<0.01 level

Table A.7: Means and Standard Deviations for Peak Kinetic Variables During a Single Leg Squat

Measure	Restricted ± SD	Control ± SD	p-Value
Hip Extension Moment	-0.032 ± 0.035	-0.032 ± 0.047	0.99
Hip Abduction Moment	-0.052 ± 0.041	0.00067 ± 0.020	<0.001**
Knee Extension Moment	-0.11 ± 0.022	-0.12 ± 0.025	0.13
Knee Varus Moment	-0.014 ± 0.035	-0.010 ± 0.025	0.65

*=Significant at p<0.05 level
**- Significant at p<0.01 level

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