The Relationship between Hip Muscular Stiffness and the Biomechanical Factors Associated with ACL Injury

Tyler Robert Cram

A thesis submitted to the faculty of the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Master of Arts in the Department of Exercise & Sport Science in the College of Arts & Sciences.

Chapel Hill
2011

Approved By:

Dr. Troy Blackburn

Dr. Darin Padua

Mr. Marc Norcross

Abstract

Tyler Robert Cram: The Relationship between Hip Muscular Stiffness and the Biomechanical Factors Associated with ACL Injury
(Under the direction of Dr. Troy Blackburn, Dr. Darin Padua, and Marc Norcross)

Anterior cruciate ligament (ACL) injury is prevalent in physically active individuals. Diminished hip neuromuscular control is thought to contribute to ACL injury. The purpose of this study was to explore the relationship between hip muscular stiffness and biomechanical ACL loading mechanisms. Subjects underwent a hip muscular stiffness assessment and a dynamic movement assessment. Individuals with greater hip abductor stiffness displayed lesser peak knee valgus angles and displacement during landing. The combination of hip extensor and abductor stiffness significantly predicted peak knee valgus angle, valgus angle at IGC, and knee valgus displacement. Our findings suggest a link between hip abductor stiffness and knee valgus motion, factors which have ACL injury implications. Because muscular stiffness can be altered, consideration should be given to the inclusion of stiffness training in ACL injury prevention programs. Future research is needed to determine the effects of increased hip muscular stiffness on ACL loading and injury risk.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

CHAPTER I

INTRODUCTION

Overview

Although injury to the anterior cruciate ligament (ACL) has been the topic of much research, it continues to be a problematic issue. There are approximately 250,000 ACL injuries occurring annually in the United States (Hewett, et al. 1999). It is estimated that one third of the 250,000 ACL injuries will undergo a reconstructive procedure resulting in surgical costs of up to $1.5 billion annually (Boden, et al. 2000b).

Most ACL injuries occur during athletic events, and of these athletic ACL injuries, it is estimated that 70% are caused by noncontact mechanisms (Griffin, et al. 2000). Noncontact ACL injury occurs most often during deceleration, lateral pivoting, or landing tasks. These tasks increase external loads applied to the knee joint (Besier, et al. 2001) resulting in increased internal force and moment requirements from active and passive structures. Although females are at a two-to-eight times greater risk for ACL injury than males (Griffin, et al. 2000), noncontact injury patterns are the same for both sexes (Ireland 1999). Ireland (1999) describes this pattern as the "position of no return" which is characterized by adduction and internal rotation of the hip, knee valgus, and external rotation of the tibia. In a prospective study by Hewett et al. (2005), peak knee valgus angle and external knee valgus moment during the loading phase of a double leg jump landing task were significant predictors of ACL injury risk. External knee valgus moments in particular had a sensitivity of 78% and specificity of 73% for predicting ACL injury risk (Hewett, et al. 2005). Additionally, participants who subsequently suffered a noncontact ACL injury had significantly greater peak vertical ground reaction forces at baseline compared to uninjured participates (Hewett, et al. 2005). These findings suggest that limiting ground reaction forces, knee valgus motion, and knee valgus moment may reduce ACL injury risk.

Vertical ground reaction forces generate external moments about the joints of the lower extremity when the foot strikes the ground. The body counteracts these external moments with its own internal moments provided by muscular contraction in order to stabilize the lower extremity. Greater vertical ground reaction forces will result in greater external moments acting on the lower extremity, thus increasing injury risk (Yu, et al. 2006). Landing in an erect posture increases vertical ground reaction forces when compared to a more flexed position (Blackburn & Padua 2009). A more erect landing posture is characterized by a more extended knee, hip, and trunk, and is associated with greater activation of the quadriceps (Chappell, et al. 2007; Quatman & Hewett 2009). The combination of shallow knee flexion angles between 10 and 40 degrees and increased quadriceps activity place excessive shear force on the ACL which can raise the possibility of rupture (Boden, et al. 2000a).

It is proposed that the body's first line of defense against joint injury is dynamic stabilization provided by the muscular structures surrounding the joint (Padua & Blackburn 2003). The degree to which dynamic stability is provided by these structures is influenced by their stiffness (Padua & Blackburn 2003). Muscle stiffness (k) is defined as the ratio of the change in force (ΔF) to the change in length (ΔL) of the muscle-tendon unit (k=ΔF/ΔL) (Blackburn, et al. 2009). Therefore, given the same applied force, a muscle exhibiting greater stiffness will undergo a smaller amount of lengthening than a muscle with lesser stiffness.

Chaudhari (2006) and colleagues demonstrated that increasing hip stiffness decreased ACL injury threshold via a frontal plane simulation model. *In vivo*, knee valgus during challenging dynamic activities have been associated with decreased neuromuscular control of the hip (Imwalle, et al. 2009; Mclean, et al. 2005). During closed-kinematic-chain activities, excessive hip adduction results in increased frontal plane knee valgus angles (Jacobs, et al. 2007), a biomechanical risk factor for ACL injury (Ireland 1999) . The inability to control frontal plane motion during landing has been linked to decreased strength and endurance of the hip abductors (Carcia, et al. 2005; Jacobs, et al. 2007). During landing and cutting the hip abductors undergo eccentric loading in order to control hip adduction. Given the same applied force, greater hip abductor muscular stiffness should result in a lesser change in length thereby decreasing hip adduction and the associated knee valgus.

Greater trunk flexion during landing results in greater peak knee and hip flexion and decreases peak ground reaction forces and quadriceps electromyography amplitude (Blackburn & Padua 2009). Conversely, landing in a more extended and erect posture increases injury risk (Ireland 1999) and may be caused by the inability to control the trunk during landing. When landing with the trunk in a more flexed position, the trunk's center of mass is farther from the hip joint axis of rotation and places a greater eccentric load on the hip extensors. Individuals with inefficient hip extensors may not be able to control trunk flexion during landing and compensate by landing with a more upright posture. Given the same applied force, greater hip extensor muscular stiffness should result in the ability to control trunk flexion during landing allowing for a more flexed landing position and decreased vertical ground reaction forces. Greater hip extensor muscular stiffness should also increase the ability to control femoral internal rotation during landing and decrease knee valgus.

Also, hip extensors such as the gluteus maximus, must also eccentrically control femoral internal rotation during landing. Since internal rotation is a dynamic component of knee valgus, weak hip extensors may increase femoral internal rotation and the associated knee valgus. Given the same applied forced, greater hip extensor muscular stiffness should allow for decreased internal rotation during landing and allowing for less knee valgus motion distally.

Greater peak vertical ground reaction forces, knee valgus angles, knee valgus moments, and lesser peak hip flexion angles during landing are biomechanical factors associated with noncontact ACL injury. Greater stiffness of the hip extensors and abductors may protect knee by limiting hip adduction and allowing for a more flexed landing posture. However, we are unaware of any research *in vivo* evaluating the influence of hip extensor and abductor muscular stiffness on biomechanical ACL injury risk factors. Therefore the purpose of our study was to explore relationships between: (1) hip abductor stiffness and knee valgus angles and moments as well as hip adduction angles and moments; (2) hip extensor stiffness and hip flexion angles, hip internal rotation angles, and vertical ground reaction forces.

Criterion Variables

Kinematics

A. Knee valgus angle
    a. Initial ground contact
    b. Peak
    c. Displacement
B. Hip adduction angle
    a. Initial ground contact
    b. Peak
    c. Displacement
C. Hip flexion angle
    a. Initial ground contact
    b. Peak
    c. Displacement

Kinetics

A. Peak internal knee varus moment
B. Peak vertical ground reaction force

Predictor Variables

Muscular Stiffness

A. Hip Abductor
B. Hip Extensor

Research Questions

RQ_1: Is there a significant relationship between hip abductor stiffness and the following variables during a jump landing task:

a. knee valgus angle at initial ground contact?
b. peak knee valgus angle?
c. knee valgus angular displacement?
d. peak internal knee varus moment?

RQ_2: Is there a significant relationship between hip abductor stiffness and the following variables during a jump landing task:

a. hip adduction angle initial ground contact?
b. peak hip adduction angle?
c. hip adduction angular displacement?
d. peak internal hip abduction moment?

RQ_3: Is there a significant relationship between hip extensor stiffness and the following variables during a jump landing task:

a. hip flexion angle at initial ground contact?
b. peak hip flexion angle?
c. hip flexion angular displacement?
d. peak vertical ground reaction force?

RQ_4: Is there a significant relationship between hip extensor stiffness and the following variables during a jump landing task:

a. internal rotation at initial ground contact?
b. peak internal rotation angle?
c. internal rotation angular displacement?

RQ_5: Does the linear combination of hip extensor stiffness and hip abductor stiffness significantly predict the following variables:

a. knee valgus angle at initial ground contact?
b. peak knee valgus angle?
c. peak knee valgus angular displacement?
d. peak internal knee varus moment?
e. peak vertical ground reaction force?

Hypothesis

Research Hypotheses

H_1: Hip abductor stiffness will be negatively correlated with the following variables during a jump landing task:

a. knee valgus angle at initial ground contact.
b. peak knee valgus angle.
c. knee valgus angular displacement.
d. peak internal knee varus moment

H_2: Hip abductor stiffness will be negatively correlated with the following variables during a jump landing task:

a. hip adduction angle at initial ground contact.
b. peak hip adduction angle.
c. hip adduction angular displacement.
d. peak internal hip abduction moment.

H_3: Hip extensor stiffness will be positively correlated with the following variables during a jump landing task:

a. hip flexion angle at initial ground contact.
b. peak hip flexion angle.
c. hip flexion angular displacement.

Hip extensor stiffness will be negatively correlated with the following variable during a jump landing task:

d. peak vertical ground reaction force.

H_4: Hip extensor stiffness will be negatively correlated with the following variables during a jump landing task:

a. internal rotation at initial ground contact.
b. peak internal rotation angle.
c. internal rotation angular displacement.

H_5: Hip extensor stiffness and hip abductor stiffness are able to significantly predict the following variables:

a. knee valgus angle at initial ground contact.
b. peak knee valgus angle.
c. peak knee valgus angular displacement.
d. peak internal knee varus moment.

Operational Definitions

Hip Abductors: Gluteus medius, gluteus minimus, and tensor fasica latae

Hip Extensors: Gluteus maximus

Dominant leg: Leg the subject would use to kick a ball for maximum distance.

Muscular Stiffness: $k = \Delta F/\Delta L$

Loading phase: Time between initial ground contact and peak knee flexion

Initial ground contact: First vertical ground reaction force measure above 10N

Jump landing task: Subjects will drop from a box with the height of 30 cm and as soon as they land they will jump straight up as high as they can.

Assumptions

1. Measure of hip abduction stiffness is an accurate measure of active hip abductor stiffness.
2. Measure of hip extension stiffness is an accurate measure of active hip extensor stiffness.
3. Subjects will perform the jump landing task to the best of their ability.

4. Muscle stiffness measured during isometric contraction is indicative of muscle stiffness during dynamic tasks such as landing.
5. A gluteus maximus that has a high stiffness when resisting hip flexion also has a high stiffness when resisting hip internal rotation.

Limitations

1. Hip muscle stiffness is novel and has no basis of comparison.
2. Comparison of a static measurement to a dynamic task.
3. Measurements will be taken in a controlled laboratory environment.
4. Limitation of the sample population.

Delimitations

1. Subjects are between ages 18-30.
2. Subjects have no prior history of lower extremity or spine surgery.
3. Subjects have not sustained a lower extremity injury in the last 6 months.

Significance

Muscular stiffness is believed to have the ability to limit excessive joint movement and therefore provide protection from injury. Greater hip abductor stiffness may be able to control knee valgus by limiting hip adduction. Also, greater hip extensor stiffness may be associated with decreased vertical ground reaction forces as well as internal rotation. For these reasons, hip muscular stiffness may allow for more control of the frontal plane motion of the knee as well as allow for a more flexed landing posture. Given that there is literature indicating that muscle stiffness can be increased through a variety of techniques, increasing stiffness may be an important component for ACL injury prevention.

CHAPTER II

REVIEW OF THE LITERATURE

Introduction

The exact etiology of non-contact ACL injury has yet to be identified. Previous research has found an association between hip musculature and the biomechanical factors associated with ACL injury. Muscular properties, such as stiffness, are believed to protect a joint against injury. Hamstring muscular stiffness may have the ability to decrease the sheer force placed on the ACL and protect it from injury (Blackburn, et al. 2004a). Since it is known that the hip musculature influences knee biomechanics (Jacobs, et al. 2007), hip muscular stiffness may be associated with the biomechanical factors that are related to knee injury. To better understand this relationship, this section will first review the anatomy of the hip, knee, and ACL. Then the epidemiology, common risk factors and potential mechanisms of non-contact ACL injury will be explored. Finally, the concept of muscular stiffness will be explained with specific emphasis on how muscular stiffness about the hip may influence biomechanics related to ACL injury.

Anatomy

The lower extremity is comprised of three major joints which include the hip, knee, and ankle. The hip joint is a very stable a ball-and-socket joint made up of the articulation between the acetabulum of the pelvis and the head of the femur. This joint is very stable because the femoral head attaches deep within the acetabulum. The hip also has a labrum which is analogous to the labrum in the shoulder and also helps to deepen the socket to increase overall joint stability (Moore & Am 2007).

Another stabilizing structure of the hip is the joint capsule. The joint capsule is formed by three ligaments – the iliofemoral ligament, the ischiofemoral ligament, and the pubofemoral ligament (Magee 2006). The iliofemoral ligament, also known as the Y ligament of Bigelow, has been considered as the strongest ligament in the body. It is responsible for preventing excessive extension. The ischiofemoral ligament attaches at the ischial aspect of the acetabular rim and runs superolaterally over the neck of the femur and then attaches just medial to the base of the greater trochanter. During extension the ischiofemoral ligament causes a wringing effect, tightening the femoral head into the acetabulum. Lastly the pubofemoral ligament connects the crest of the pubic bone and the base of the iliofemoral ligament on the femur. This ligament prevents excessive abduction and limits extension.

Since the hip is a ball-and-socket joint it is able to move about all three planes: sagittal, frontal and tranverse. In the sagittal plane the hip is able to undergo flexion and extension. The musculature that is responsible for extension is located on the posterior thigh. The prime hip extensors are the gluteus maximus and the posterior part of the gluteus medius. When the knee is locked in extension the hamstring muscle group assists in hip extension. As the knee flexes, the hamstrings contribution to hip extension decreases due to the active insufficiency of the muscle group. These muscles are the biceps femoris, semitendinosus, and semimembranosus. The major hip flexor is the iliopsoas which is comprised of the psoas major and the iliacus. Accessory hip flexors are the retus femoris, sartorius, tensor fascia latae, and pectineus (Kendall, et al. 2005).

In the fontal plane there are several muscles that abduct and adduct the hip. The muscles responsible for hip abduction are the gluteus medius and gluteus minimus. The tensor fascia latae and upper fibers of the gluteus maximus assist in hip abduction. The muscles responsible for hip adduction include the adductor longus, adductor brevis, adductor magnus, pectineus, and gracilis(Kendall, et al. 2005).

In the transverse plane the hip internally and externally rotates. Internal rotation musculature includes the gluteus medius, gluteus minimus, and tensor fascia latae. The adductor longus is also considered to assist in hip internal rotation. The hip external rotators are located under the gluteus maximus on the posterior thigh and are comprised the piriformis, obturator externus, gemellus superior, quadrates femoris, gemellus inferior, and obturator internus. When the hip is extended the gluteus maximus assists in external rotation(Kendall, et al. 2005).

The next joint of importance distal to the hip joint is the knee, or the tibiofemoral joint. The tibiofemoral joint is a modified hinge joint with two degrees of freedom and is considered the largest joint in the body (Moore & Am 2007). It is comprised of two articulations – the medial and lateral femorotibial articulations. The knee is surrounded by a synovium that almost encapsulates the entire joint (Duthon, et al. 2006). The interior of the knee houses the menisci and the cruciate ligamens. The menisci are split into a medial c-shaped structure and a lateral o-shaped structure. These structures are made of fibrocartilage and lie on the articular surface of the tibia. The major role of the menisci is to deepen the surface of the articulation between the femur and the tibia as well as to provide shock absorption (Moore & Am 2007).

The knee is surrounded by several extracapsular ligaments. These ligaments are the lateral collateral ligament, the medial collateral ligament, and the arcuate ligament. The lateral collateral ligament is a cord like structure that runs from the lateral epicondyle of the femur to the lateral side of the fibular head. The medial collateral ligament is a flat band structure that runs from the medial epicondyle of the femur to the medial condyle of the tibia. The deep fibers of the medial collateral ligament have an attachment to the medial meniscus. The arcuate ligament is part of the posterior lateral corner and strengthens the posterolateral portion of the capsule. It runs from the posterior aspect of the fibular head and spreads over the posterior surface of the knee joint.

The major intraarticular ligaments of the knee are the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL). The ACL attaches on the anterior intercondylar area of the tibia and runs superiorly and posterolaterally to insert on the posterior aspect of the medial side of the lateral condyle of the femur. The ACL plays a crucial role in joint stability (Duthon, et al. 2006) because it is the primary restraint to anterior translation of the tibia relative to the femur (Duthon, et al. 2006; Zantop, et al. 2006). The PCL attaches on the posterior aspect of the tibia and runs superiorly and anterieromedially to insert on the femoral intercondylar notch. Contrary to the ACL, the PCL is the primary restraint to posterior translation of the tibia relative to the femur (Amis, et al. 2006). Although the knee has a synovium that almost encapsulates the entire joint, the distribution of synovium leaves the cruciate ligaments extrasynovial (Duthon, et al. 2006).

Based on fiber function, the ACL can be divided into two parts – the anteromedial bundle, which is more taught in 20-90 degrees of flexion, and the posterolateral bundle is more taught near full extension (Duthon, et al. 2006; Zantop, et al. 2006). The ACL is also a major secondary restraint against femoral internal rotation. The posterolateral bundle has been suggested to be more responsible for rotational stability compared to the anteromedial bundle (Zantop, et al. 2006).

The two major muscle groups that act on the knee are the quadriceps and the hamstrings. The quadriceps insert on the patella and work together to extend the knee. The quadriceps muscles are the recuts femoris, vastus lateralis vastus medialis, and the vastus intermedius. The hamstrings are responsible for flexing the knee, extending the hip when the knee is in extension and producing internal and external rotation of the tibia on the femur. The hamstrings are comprised of the semitendinosus, semimembranosus, and the biceps femoris (Kendall, et al. 2005).

Epidemiology

The ACL is one of the most commonly injured ligaments of the knee (Boden, et al. 2000b). There are approximately 250,000 ACL injuries - 1 in every 3,000 people - annually in the United States (Hewett, et al. 1999). It is estimated that one third of the 250,000 ACL injuries will undergo a reconstructive procedure resulting in surgical costs of up to $1.5 billion annually (Boden, et al. 2000b). This annual cost does not include the cost of those who choose nonsurgical treatment and care or the long term care of these ACL deficient patients (Boden, et al. 2000b).

Knee osteoarthritis is the greatest cause of disability in the United States (Palmieri-Smith & Thomas 2009). Knee joint injuries, including ACL rupture, increase the risk of developing osteoarthritis (Gelber, et al. 2000) compared to those who do not sustain injuries. It is estimated that sustaining an ACL rupture leads to osteoarthritis approximately 5-20 years post injury (Palmieri-Smith & Thomas 2009). Current literature suggests that there is no difference in incidence of osteoarthritis in those that receive ACL reconstructions and those who do not (Lohmander, et al. 2007). Since most ACL injuries occur during athletics in the teenage population, osteoarthritis can present in adults as early as their 20s and 30s (Gelber, et al. 2000). For this reason emphasis must be kept on prevention.

Males sustain a higher number of ACL injuries than females because there are a greater number of males who participate in sports (Hewett, et al. 2005). However, since the passage of Title IX there has been a large increase in female sport participation especially at the high school level (Hewett, et al. 2005). In sports that have comparable equipment and rules such as soccer, basketball, and volleyball there are a greater total number of injuries in males, however, a greater relative number of females sustain ACL injuries. (Griffin, et al. 2000).

Mechanisms

ACL injury results from one of two categories of injury mechanisms – contact or noncontact. It is estimated that approximately 30% of all injuries to the ACL are due to direct contact by another player or object which can occur in sports such as skiing or football (Griffin, et al. 2000). The remaining 70% of ACL injuries are non-contact in nature. Video analysis suggests there are two different movement patterns that load the ACL and lead to rupture: knee valgus collapse and anterior tibial shear (Quatman & Hewett 2009). Ireland suggests a similar injuring movement pattern as the 'position of no return' (Ireland 1999). This position is comprised of hip internal rotation and adduction, knee valgus and tibial external rotation on a pronated and externally rotated foot. Since loading of the ACL can occur in the frontal, sagittal, and transverse planes it is believed that there is not one plane that completely accounts for the cause of rupture, but rather that it is a multiplanar phenomenon (Quatman & Hewett 2009).

Risk Factors

There are several risk factors that have been associated with noncontact ACL injury. These risk factors can be grouped into extrinsic and intrinsic risk factors. Extrinsic risk factors are those that do not originate from the body. These include surface type and shoe-turf interface. Different shoes have varying coefficients of friction which can influence joint kinematics (Boden, et al. 2000a). A 5 year prospective study by Meyers & Bamhill (2004) revealed a higher incidence of injury on FieldTurf vs. natural grass. Intrinsic risk factors that are of the most concern are hormonal, anatomic, biomechanical, and neuromuscular.

<u>Hormonal</u>

The levels of circulating hormones in the body have been a recent area of investigation in the female athlete (Bell, et al. 2009). Hormones such as estrogen, progesterone, and relaxin have been shown to have an influence on the integrity of the ligament at different times during menses. On the ACL there are receptor sites for these hormones which can affect ligamentous laxity, fibroblast proliferation, and collagen synthesis (Childs 2002). Shultz et al found that hormonal concentrations of testosterone estrogen, and progesterone explained 63% of the variance in anterior tibial translation during the menstrual cycle (Shultz, et al. 2004).

Anatomical

ACL size, intercondylar width, joint laxity, Q-angle, and pelvis size are several anatomic intrinsic risk factors for ACL injury. Decreased interconylar notch width and increased ACL size has been shown to increase ACL injury risk because it becomes impinged against the notch resulting in a shear force being applied to the ligament (Boden, et al. 2000b). A wider pelvis increases an individual's Q-angle. The Q-angle is the measurement of the angle between a line drawn from the ASIS to the midpoint of the patella and a line drawn from the tibial tuberosity and the midpoint of the patella (Magee 2006). Normal Q-angle for males is 13° and normal for females is 18°(Magee 2006). An increased Q-angle places the knee into a more valgus position placing stress and on the medial knee structures as well as the ACL.

Neuromuscular

Of the intrinsic risk factors discussed, hormonal and anatomic factors are very difficult to alter. However neuromuscular risk factors leave the greatest room for intervention. Noncontact ACL injury occurs most often during deceleration, lateral pivoting, or landing tasks which increase external loads applied to the knee joint (Besier, et al. 2001). The exact etiology of non-contact ACL injury is still unknown. However, an enormous amount of studies have investigated the movement patterns that may lead to insult (Blackburn, et al. 2009; Blackburn & Padua 2008; Blackburn & Padua 2009; Carcia, et al. 2005; Gehring, et al. 2009; Hanson, et al. 2008; Hewett, et al. 2005; Hewett, et al. 2009; Imwalle, et al. 2009; Jacobs, et al. 2007; Mclean, et al. 2004; Mclean, et al. 2005; Nyland, et al. 2004; Padua, et al. 2009; Quatman & Hewett 2009; Tsai, et al. 2009; Wallace, et al. 2008). Not surprisingly these movement patterns incorporate many of the characteristics described by the "position of no return" including knee valgus angles and moments, vertical ground reaction force, and upright landing postures.

Knee Valgus. Dynamic knee valgus occurs with hip adduction and internal rotation while the knee undergoes adduction (Ireland 1999). This position can load the ACL in the frontal, sagittal, and transverse planes, potentially and is why it has the ability to cause rupture (Quatman & Hewett 2009). Hewett et al. (2005) found that knee valgus angles and moments predict ACL injury risk.

Hewett prescreened 205 adolescent female athletes prior to the start of their season using 3D biomechanical analysis of a drop vertical jump from a 30cm high box. Nine subjects went on to sustain an ACL injury diagnosed by arthroscopic surgery or MRI. Knee abduction angle, knee abduction moment, and vertical ground reaction force were statistically significantly greater in those athletes that went on to rupture their ACLs. Knee abduction angle and abduction moments were able to predict ACL injury. Although these findings are valuable, the sample size was very small, thus it is unclear how generalizable the findings are to the population at large.

Although the exact reason why some people exhibit higher knee valgus during dynamic activity is still unknown, many researchers feel that inefficient hip musculature may lead to greater knee valgus. During knee valgus the hip goes into adduction and internal rotation. In order to decrease the amount of hip adduction the hip abductors, such as the gluteus medius, must become activated (Carcia, et al. 2005). Hip musculature with greater gluteal strength and activation may be able to control hip internal rotation and femoral adduction limiting the associated knee valgus.

Several studies have evaluated hip abductor fatigue and its influence on knee valgus angles. (Carcia, et al. 2005; Gehring, et al. 2009; Jacobs, et al. 2007; Kernozek, et al. 2008; Mclean, et al. 2005) These studies found that subjects with greater hip abduction strength exhibit less knee valgus when landing from a jump or cutting task. These studies also show that females experience greater difference in landing kinematics after fatigue. Jacobs et al (Jacobs, et al. 2007) found that males demonstrated a small positive correlation between hip abduction strength and landing kinematics whereas females demosntrated a larger negative correlation between abduction strength and landing kinematics. This suggests that hip abductor strength plays a more substantial role in controlling landing kinematics in females than in males and that insufficient hip abduction strength is associated with greater knee valgus when landing from a jump.

Imwalle et al. (2009) investigated hip and knee kinematics during different cutting angles in females. They found that knee abduction angles during a 90 ° cut and 45° cut was predicted by hip adduction angle. Their findings also suggested that hip transverse plane kinematics were not associated with knee abduction angles. Strategies to prevent knee abduction should be focused on neuromuscular control at the hip.

Erect landing posture & ground reaction force. Along with knee abduction, hip sagittal plane kinematics during landing have also been linked with ACL injury. Landing from a jump in a more erect posture has been associated with higher vertical ground reaction forces (Blackburn & Padua 2009). More erect landing postures are a result of a more extended knee, hip, and trunk. Higher vertical ground reaction forces have been linked to greater ACL injury risk (Hewett, et al. 2005). In order to come to a stop during a landing the quadriceps must produce an eccentric contraction to counter act the external knee flexion moment caused by the vertical ground reaction force. With an increased vertical ground reaction force there is an increase in quadriceps activity. Quadriceps activity, especially at lower knee flexion angles, cause an anterior tibial shear force loading the ACL(Padua, et al. 2009). For this reason landing more upright from a jump puts one at risk for injury to their ACL.

Blackburn et al (Blackburn & Padua 2009) compared preferred landing strategies to active trunk flexion during landing. When subjects landed with active trunk flexion they displayed increased hip and knee flexion angles as well as decreased vertical ground reaction force while exhibiting less activation of the quadriceps.

People may land in a more erect posture due to weak and inefficient hip extensors. When landing from a jump with a more flexed posture the center of mass is farther away from the axis of rotation at the hip which places a greater eccentric load on the hip extensors. In order to prevent their body from collapsing in this position, the hip extensors must be able to handle this load. Although a more flexed landing posture decreases vertical ground reaction forces and risk of injury, if the hip extensors are weak and unable to control the load the body may land in a more up right and erect posture.

Yu et al (Yu, et al. 2006) investigated hip and knee kinematics during a stop vertical jump task and the associated relationship with vertical ground reaction forces. The results of this investigation show that hip and knee joint flexion and extension angular velocities at initial ground contact have an effect on vertical ground reaction force where as hip and knee flexion angles did not. This suggests that when landing, it is active motion at the knee and the hip that helps decrease vertical ground reaction force rather than the angular position. Therefore, if greater hip and knee flexion moments decrease vertical ground reaction force during landing, active flexion during landing may lead to a decrease in the activation of the quadriceps and ultimately decrease the load placed on the ACL.

Whether the sagittal plane anterior tibial shear force or the frontal plane knee valgus is the major culprit of ACL injury is still up for debate (Blackburn & Padua 2008; Mclean, et al. 2005; Quatman & Hewett 2009). Some researchers feel that the frontal plane motion exhibited at the knee is the major mechanism of injury and sagittal plane motion is insignificant (Mclean, et al. 2005; Quatman & Hewett 2009). Other researchers feel that motion in the sagittal plane, especially exhibited in the quadriceps dominant females, increases the risk for injury (Chappell, et al. 2002). Rather than one plane being solely responsible for ACL injury, Blackburn (2008) suggests that frontal and transverse plane motions preload the ACL that when coupled with a large magnitude quadriceps contraction causes a rupture of the ligament.

Muscle Stiffness

It is believed the body's first line of defense from injury comes from the material stiffness properties of a joint's surrounding muscles and soft tissue structures (Padua & Blackburn 2003). Muscle stiffness should not to be confused with muscle range of motion. Muscle range of motion does not take into account the muscle's resistance to the change in length. Muscle stiffness is defined by the equation $K = \frac{\Delta F}{\Delta L}$ where K is the stiffness of the muscle, ΔF is the change in force and ΔL is the change in length (Blackburn, et al. 2009). When comparing muscle stiffness, a stiffer muscle would demonstrate a greater increase in force for the same change in length compared to a more compliant/less stiff muscle. Therefore, given the same force application, a stiffer muscle remains in a relatively shorter position and thus would allow less joint motion in the antagonistic direction. This decreased motion may decrease the load transferred to ligamentous structures.

Increased active knee flexor stiffness has been shown to be associated with decreased anterior tibial translation protecting the ACL from injury (Blackburn, et al. 2009; Blackburn, et al. 2004b). Blackburn et al. reported that females had greater active extensibility but males had greater active and passive stiffness of the hamstrings. This may contribute to the higher rate of ACL injury in females. Due to the decreased knee flexor stiffness females are unable to meet the high loads placed on the lower extremity during physical activity leaving the ACL at risk for injury.

To date there have not been any investigations that have looked into hip muscular stiffness in vivo, however Chaudhari (2006) was able to create a simulation model that evaluated the influcence of muscular stiffness at the hip and ankle on ACL injury threshold. In this study they created a three-link frontal plane model of a support limb where they were able to measure the maximal sustainable axial force the limb could sustain before the joint opened medially or laterally by 8° which was used as the ACL injury threshold. They found that an increase in hip stiffness by 50% also increased injury threshold from 5.1 times body weight to 5.4 times body weight with neutral alignment and nominal hip stiffness. The injury threshold dropped drastically with increasing valgus or varus alignment. Although this study used simulation models, it proves a good rationale for an investigation looking at hip muscular stiffness' influence on ACL injury risk factors.

Being able to identify those with decreased muscle stiffness may be important considering its potential contribution to joint stability. Kubo (Kubo, et al. 2006a; Kubo, et al. 2006b) investigated the effects of isometric training on changing muscle stiffness. Subjects completed a 12 week unilateral isometric knee extension training program on a dynamometer at 70% of their maximal contraction. Right and left legs were randomized to a shortened position 50° and the other leg in a more lengthened position at 100°. Results showed that there was a significant increase in stiffness of the tendon structures for the 100° condition but not for the 50° condition.

Kubo et al. (2006b) also investigated the effect of isometric squat training on tendon stiffness and squat performance. A total of 14 subjects volunteered for the study. Eight subjects were put in a training ground and the other 6 subjects were used as controls. The training group completed training sessions 4 times per week for 12 weeks. Training sessions included 10 repetitions of 70% maximum voluntary contraction for 15 seconds on a horizontal leg press. Results showed that the training group significantly increased their knee extensor stiffness compared to the control group. These studies show that stiffness of the muscle tendon complex can be altered with isometric training.

Conclusion

As demonstrated by this review of the current literature, hip adduction during landing leads to knee valgus and erect landing postures lead to higher vertical ground reaction forces which have been associated with those who get injured. Hip adduction maybe be controlled by greater hip abductor stiffness and greater hip extensor stiffness may allow for a more flexed landing posture. In order to evaluate this relationship I am going to compare hip abductor stiffness and hip extensor stiffness to knee valgus angles and moments, femoral internal rotation, hip flexion angle, and vertical ground reaction force.

CHAPTER III

METHODOLOGY

Subjects

A convenience sample of 40 healthy subjects (20 males and 20 females; age = 20.20 ± 1.63 years, height = 173.42 ± 9.38 cm, mass = 71.41 ± 14.52 kg) volunteered to participate in this investigation. Subjects were recreationally active, defined as participating in at least 30 minutes of exercise a minimum of 3 times per week, and between the ages of 18-30 years. Subjects were excluded if they sustained any injury to the lower extremity in the 6 months prior to data collection that limited their activity for at least 3 consecutive days, or had a history of chronic lower extremity injury, lower extremity or spine surgery, or neurological disorder.

Study Design

Subjects to reported to the Neuromuscular Research Laboratory for one testing session. Subjects first were prepared for electromyography (EMG) electrode placement and then completed a set of maximum voluntary isometric contractions (MVICs) for each muscle group. Following the MVIC assessments, subjects completed a muscular stiffness assessment and a dynamic movement assessment in a counterbalanced order. Details of these assessments are described below.

Procedures

All subjects read and signed an informed consent document approved by the Institutional Review Board at the University of North Carolina at Chapel Hill prior to data collection. Height and mass were measured and the dominant leg was established, defined as the leg they would use to kick a ball for maximum distance. Next, subjects were prepared for EMG measurements. EMG data was recorded in order to monitor muscle activation during the dynamic task. Preamplified/active surface EMG electrodes (DelSys Inc., Boston, MA: interelectrode distance = 10 mm; amplification factor = 10,000 (20–450 Hz); CMMR at 60 Hz > 80 dB; input impedance > 10^{15} Ω/0.2 pF) were placed over the muscle bellies of the gluteus maximus (GMax), gluteus medius (GMed), medial hamstrings (MHam), and biceps femoris long head (LHam) parallel to the direction of action potential propagation. Electrodes were placed between innervation zones and distal attachments as described by Rainoldi et al.(2004). Correct electrode placements were verified using an oscilloscope during manual muscle testing (Kendall, et al. 2005).

Maximal Voluntary Isometric Contraction Assessment

Following electrode placement, all subjects completed the maximum voluntary isometric contraction (MVIC) assessment with the order of the muscle group being tested counterbalanced.

Hip Extension: Subjects were positioned prone on a table with the anterior iliac crest (ASIS) located at the end of the table, the hip positioned in neutral, and the knee in 90° of flexion. Subjects were instructed to extend the hip with maximal effort while resistance was applied just proximal to the popliteal fossa on the posterior aspect of the thigh with a hand-held dynamometer (Lafayette instrument, Lafayette, Indiana).

Hip Abduction: Subjects were positioned sidelying on the table on the non-dominant side and the dominant leg kept in full extension. Subjects were instructed to abduct the hip with maximal effort while resistance was applied just proximal to the joint line of the knee with a hand-held dynamometer.

Knee Flexion: Subjects were positioned prone on the table with the knee at 90° of flexion and the shank perpendicular to the floor. Subjects were instructed to flex the knee with maximal effort while resistance was applied to the distal portion of the shank with a hand-held dynamometer.

Subjects completed three 5 second trials for each hip motion and knee motion during which EMG amplitudes were sampled. Norcross et al.(Norcross, et al. 2009) demonstrated high intrasession reliability for EMG amplitudes of the GMax and GMed using these procedures ($ICC_{2,1}$ =0.95, SEM= 49.4mV and $ICC_{2,1}$ 0.98, SEM = 50.4mV, respectively). Peak forces measured during the MVIC trials using the handheld dynamometer were recorded. These results were averaged and used for loading during muscular stiffness assessment which is calculated at 30% of MVIC. Following MVIC assessments, subjects then completed a muscular stiffness assessment and a dynamic movement assessment with the order of these assessments counterbalanced for all subjects.

Muscular Stiffness Assessment

Muscular stiffness was calculated by modeling the hip as a single degree of freedom mass spring system and observing the damping effect of the hip extensors and hip abductors on oscillatory hip motion during an unanticipated perturbation (Blackburn, et al. 2004b). Five trials were completed for both the hip extensors and abductors with 1 minute of rest between trials to minimize the likelihood of fatigue. Good intrasession reliability was found for both hip extensor ($ICC_{2,1}$= 0.83, SEM = 551.7 Nm) and hip abductor stiffness ($ICC_{2,1}$= 0.74, SEM = 800.46 Nm) during pilot testing. The order of the stiffness conditions were counterbalanced across subjects.

Hip Extensor Stiffness: Subjects were positioned prone on a table with the ASIS located at the edge of the table. The hip was maintained in neutral and the testing knee was locked in 90° of flexion in a post-operative knee brace (Bledsoe Brace Systems, Grand Prairie, TX). An accelerometer (PCB Piezotronics, Depew, NY) was placed on the lateral hinge of the knee brace and a load equaling 30% of the hip extension MVIC force was placed on the posterior aspect of the thigh just proximal to the popliteal fossa using standard cuff weights as seen in figure 5. The thigh was then placed parallel to the floor and subjects were instructed to contract the hip extensors isometrically to support the thigh in this position. The investigator applied a downward perturbation to the posterior thigh forcing the hip into flexion causing an oscillatory motion. This was done at random within 5 seconds during which accelerometer and surface EMG data were sampled.

Hip Abductor Stiffness: Subjects were positioned sidelying on the non-dominant side with the dominant knee locked in extension using a post operative brace and the hip was maintained in neutral. An accelerometer was placed on the lateral hinge of the knee brace and a load equaling 30% of the hip abduction MVIC force was placed on the lateral aspect of the thigh just proximal to the lateral knee joint line using standard cuff weights. The test leg was positioned parallel to the floor and subjects were instructed to activate the hip abductors isometrically to maintain this position. The investigator then applied a downward perturbation to the lateral thigh forcing the hip into adduction causing an oscillatory motion. This was done at random within 5 seconds during which accelerometer and surface EMG data were sampled.

Dynamic Movement Assessment

A jump landing task was used for the dynamic movement assessment. Electromagnetic motion capture sensors (Motion Star; Ascension Technologies Inc, Burlington, VT) were secured on the pelvis, shank, and thigh of the dominant leg using double sided tape, pre-wrap, and white athletic tape. The world and segment axis systems were defined using the right hand rule where positive x was forward/anterior, positive y was leftward/medial, and positive z was upward/superior (Blackburn & Padua 2009). Once the coordinate systems were established and the sensors were secured to the segments, the boney landmarks were digitized to determent segment endpoints and joint centers. The joint centers of the knee and ankle were estimated as the midpoint between the medial and lateral malleoli and the medial and lateral femoral condyles respectively.

The pelvis joint center was calculated using the bell method and digitized using the right and left ASIS (Bell, et al. 1990).

Jump Landing Task (Padua, et al. 2009): Sujects jumped from a 30 cm box to a force plate (model 4060-NC; Bertec Corp, Columbus, OH) that was located a distance equal to 50% of the participants height in front of the box. Immediately after landing, subjects completed a vertical jump for maximum height. Subjects were instructed to land on both feet, keeping the dominant foot on the force plate and to focus on gaining maximum vertical height after landing. Subjects were allowed as many practice trials as needed to feel comfortable with the task. A total of 5 successful trials were completed for each subject with 1 minute of rest between trials to reduce the likelihood of fatigue. For a successful trial, subjects had to jump off the box with both feet, land with their dominant foot completely on the force plate and their non-dominant foot completely off force plate. If these criteria were not met the trial was repeated.

Data Sampling and Reduction

Electromagnetic sensor coordinate data were sampled at 100 Hz, while force plate, accelerometer, and EMG data were sampled at 1,000 Hz. Kinematic data were time synchronized with EMG, kinetic, and accelerometer data and re-sampled at 1,000 Hz via linear interpolation. All data were captured and synchronized using The MotionMonitor software (Innovative Sports Training: Chicago, IL).

EMG data were corrected for DC bias; bandpass (20-350Hz) and notch (59.5-60.5Hz) filtered; then smoothed using a 25 ms root mean square sliding window function (Norcross, et al. 2009). All EMG data were normalized to mean activation amplitude during the middle three seconds of the respective GMax/GMed/MHam/LHam MVIC trials.

Accelerometer data were low pass filtered at 10Hz with a fourth order zero phase lag Butterworth filter. The first two oscillatory peaks in the tangential leg acceleration profile were identified and used to calculate the damped frequency of oscillation $\frac{1}{t_2 - t_1}$. This value was then be used to calculate the linear stiffness (Blackburn, et al. 2004b)using the equation $k = 4\pi^2 m f^2$, where k is linear stiffness, m is the summed mass of the system (thigh, shank, foot, brace and the applied load), and f is the damped frequency of oscillation. Because stiffness increases as a function of mass (Blackburn, et al. 2009), these data were normalized to body mass prior to statistical analyses.

Kinematic data were lowpass filtered at 10 Hz with a 4th order zero phase lag Butterworth filter. Euler angles were used for all kinematic calculations and were calculated in an YX'Z" flexion/extension, abduction/adduction, and internal/external rotation sequence. Knee angles were calculated as the shank relative to the thigh and hip angles were calculated as the thigh relative to pelvis. Kinetic data were low-pass filtered at 50Hz with a fourth order zero phase lag Butterworth filter and combined with kinematics via an inverse dynamics solution (Gagnon & Gagnon 1992) to calculate internal frontal plane knee moments.

Peak knee valgus angle, internal knee varus moment, hip flexion angle, and vertical ground reaction force were calculated during the loading phase of the jump landing task. Frontal plane knee and sagittal plane hip angles at initial contact and total displacement during the loading phase were also calculated. The loading phase was defined as the interval from initial ground contact (vertical ground reaction force > 10 N) to peak knee flexion. Ground reaction forces were normalized to body mass ($N \cdot kg^{-1}$) and moments were normalized to the product of the subject's mass and height.

Statistical Analysis

All statistical analyses were performed using SPSS version 18.0. Simple Pearson correlations were used to assess research questions 1, 2, 3, and 4 which are found in **Table 1**. Research question 5, also found in **Table 1,** was assessed using multiple regression analysis with the order of entry beginning with the highest simple correlation. Statistical significance will be established a priori as $\alpha \leq 0.05$.

CHAPTER IV

RESULTS

Hip Abductor Stiffness

While the relationship between hip abductor stiffness and the knee valgus angle at initial ground contact was non-significant (r = -0.063, P = 0.350), hip abductor stiffness was significantly correlated with peak valgus angles (r = 0.266, P = 0.048) as well as valgus angular displacement (r = 0.370, P = 0.009). Knee valgus was given a negative angular convention, therefore these data indicate that subjects with greater hip abductor stiffness displayed lesser peak knee valgus angles and valgus displacement. The relationship between hip abductor stiffness and the peak internal knee varus moment was non-significant (r = -0.153, P = 0.173). Hip abductor stiffness was significantly correlated with hip adduction angles at initial ground contact (r = 0.458, P = 0.001), peak hip adduction angle (r = 0.480, P = 0.001), and peak internal hip abduction moment (r = 0.291, P = 0.034). Hip adduction was given a positive angular convention, therefore these results indicate that subjects with greater hip abductor stiffness landed with greater hip adduction at initial ground contact, and displayed greater peak hip adduction angles and lesser internal hip abduction moment compared to those with less stiff/more compliant hip abductors. The relationship between abductor stiffness and hip adduction angular displacement was not significant (r = 0.051, P = 0.378). Complete hip abdcutor simple correlation results can be found in table #3 and table #4.

Hip Extensor Stiffness

Hip extensor stiffness was not significantly correlated with hip flexion at initial ground contact (r = 0.112, P = 0.246), peak hip flexion (r = -0.096, P = 0.278), hip flexion angular displacement (r = -0.174, P = 0.142), or peak vertical ground reaction force (r = -0.184, P = 0.128). Additionally, hip extensor stiffness was not significantly related to hip internal rotation at initial ground contact (r = 0.063, P = 0.350), peak hip internal rotation (r = 0.105, P = 0.260), or hip internal rotation angular displacement (r = 0.091, P = 0.289). Complete extensor stiffness simple correlation results can be found in table #5 and table #6.

Multiple Regression

Using multiple regression analysis, we found that the linear combination of hip abductor stiffness and extensor stiffness was significantly related to knee valgus angle at initial ground contact (R^2 = .360, P = < 0.001), peak knee valgus angle (R^2 = 0.154, P = 0.045), and knee valgus displacement (R^2 = 0.298, P = 0.001). The linear combination of abductor stiffness and extensor stiffness was not correlated with peak internal knee varus moment (R^2 = 0.062, P = 0.292) or vertical ground reaction force (R^2 = 0.060, P = 0.321). Complete multiple regression results and equation models can be found in table #7.

CHAPTER V

DISCUSSION

The primary findings of this study showed that individuals with greater hip abductor stiffness displayed less knee valgus motion during a dynamic movement task. Also, individuals with greater hip abductor stiffness displayed greater peak hip adduction during the dynamic task. Hip extensor stiffness did not significantly correlate with any of the selected biomechanical variables. While in isolation, the predicative ability for abductor stiffness and valgus at initial contact, peak valgus, and valgus displacement was non-significant, 7%, and 14% respectively. However, when hip abductor stiffness is combined linearly with hip extensor stiffness, its predictive ability increases for valgus at intitial contact, peak valgus, and valgus displacement to 36%, 15%, and 30% respectively. These results suggest that hip abductor muscular stiffness has the ability to influence landing biomechanics related to ACL injury risk.

To our knowledge this is the first investigation to evaluate relationships between hip muscular stiffness and landing biomechanics related to ACL injury. Therefore, comparison to previous literature is limited. However, our findings are consistent with previous literature supporting a relationship between hip neuromuscular control and biomechanical factors that are linked with knee injury (Carcia, et al. 2005; Gehring, et al. 2009; Jacobs, et al. 2007).

Fatigued muscles perform in a similar fashion as weak muscles due to altered force production (Millet & Lepers 2004). There have been several studies investigating the effect of hip abductor fatigue on landing mechanics to evaluate their influence on frontal plane ACL injury risk factors. Jacobs (2007) and Gehring (2009) evaluated gender differences of knee biomechanics following a hip abductor fatigue protocol. Both studies found that when compared to males females exhibited greater peak knee valgus during the post-fatigue landing task potentially increasing their risk for ACL injury. Using a similar study design, Carcia (2005) also investigated the effect of hip abductor fatigue on landing biomechanics. Contrary to Jacobs and Gehring and similar to our investigation, Carcia found greater knee valgus after a fatigue protocol inducing decreased hip abductor neuromuscular control regardless of gender. They found that subjects landed with more knee valgus at initial ground contact following fatigue protocol. However, they did not find a relationship between peak valgus or valgus displacement which may have more influence on ACL injury compared to valgus angles at initial ground contact.

Our findings were able to confirm Chaudhari's findings *in vivo* where he used a simulation model to show a link in between increased hip muscular stiffness and increased injury threshold . We were also able to confirm our primary hypothesis that stiffer hip abductors are associated with lesser knee valgus motion. The rationale behind our hypothesis was that greater hip abductor stiffness provides greater resistance to lengthening, thus limiting hip adduction motion distally. Since hip adduction contributes to dynamic knee valgus in the frontal plane, limiting this motion will also limit the amount of valgus the knee undergoes during dynamic activities. However, we found that precisely the opposite occurred: individuals with greater hip abductor stiffness landed in a less abducted position. The explanation for this unexpected finding is not completely clear. Landing in a more adducted position may be a protective landing strategy to mechanically assist the hip abductors to control knee valgus and knee valgus displacement. Although, this is most likely not the case because the hip abductors have been shown to exhibit linear increases in strength as they move from an abducted position through neutral and into a slightly adducted position (Ryser, et al. 1988).

If individuals that have decreased hip abductor stiffness display greater knee valgus displacement, they must land in a position that will allow them to undergo enough knee valgus without causing injury. One way they may accomplish this is by landing in a more hip abducted position to allow for total knee valgus motion. This may represent a phenomenon already described in the literature know as ligament dominance which occurs when individuals allow knee ligaments rather than lower extremity musculature to absorb ground reaction forces. (Myer, et al. 2004) Ligament dominant landing preference is typically characterized by increased medial knee displacement as well as high ground reaction forces (Ford, et al. 2003). However, we did not find any relationship between hip neuromuscular control and increased ground reaction forces.

In order to better understand our results we organized the data into tertiles based on hip abductor stiffness and ran secondary analysis. We then used descriptive statistics and independent samples t-tests to analyze biomechanical data of the high stiffness group (T1) and the low stiffness group (T3). Results from this secondary analysis were able to show both the high stiffness group and the low stiffness group went through the same amount of hip adduction displacement. Although we did not include foot positioning in our motion capture model, it is a possibility that individuals with lesser hip abduction stiffness landed with their feet further apart. If those who displayed lesser hip abduction stiffness truly landed in a wider stance it would explain why they displayed more hip abduction at initial ground contact and displayed lesser peak hip adduction angles. Also with their feet further from the midline when landing, they would also exhibit greater peak knee valgus and knee valgus displacement for the same amount of hip adduction displacement.

It may also be plausible that subjects with lesser hip abductor stiffness activated their hip abductors more in preparation for landing resulting in a more abduction hip at initial ground contact. To further evaluate this possible relationship we ran simple correlations between gluteus medius muscle activation during the jump landing task and hip adduction angles at IGC, peak adduction angle, and hip adduction displacement. Our findings from this analysis did not reveal any significant relationships. Future research needs to investigate other potential reasons to explain why those with increased muscular hip stiffness land in a more hip adducted position than those with less hip abductor stiffness.

Hip extensor stiffness was not correlated with peak hip flexion angle, internal rotation angle, or vertical ground reaction force, factors which have been previously associated with ACL injury risk (Blackburn & Padua 2009; Hewett, et al. 2005). For the purposes of our investigation we viewed the gluteus maximus as the primary hip extensor. The gluteus maximus assists in external rotation, but it is not the only hip external rotator. For this reason its stiffness alone may not adequately represent hip external rotator stiffness which could explain the lack of correlation with peak hip internal rotation during the task.

While the jump landing task may have been difficult enough to show neuromuscular deficits in the frontal plane, this task may not been complex enough to replicate realistic functional demands to reveal any relationships with hip extensor stiffness and associated injury risk factors. Blackburn (2009) was able to show that increased trunk flexion during from a box height of 60cm was able to decrease vertical ground reaction forces. Since vertical ground reaction forces during jump landing tasks increase as box height increases (Yeow, et al. 2009) we may have needed to use a higher box greater than 30cm in order to find differences in hip extensor stiffness particularly in the sagittal plane. Nevertheless, we chose this task for our investigation because it has been used previously (Hewett, et al. 2005) to evaluate and predict ACL injury risk.

Although hip extensor stiffness was not correlated with landing biomechanics in isolation, the linear combination of hip abductor and extensor stiffness predicted 36% of the variance in knee valgus angle at ground contact, 15.4% of the variance in peak knee valgus angle, and 29.8% of the variance in knee valgus displacement. While on the other hand, hip abductor stiffness was only able to significantly predict 7% of peak knee valgus and 14% of knee valgus displacement. Dynamic knee valgus is a multiplanar motion which consists of femoral adduction, femoral and tibial rotation. Since dynamic knee valgus occurs in multiple planes, it is understandable why the combination of stiffness in the frontal plane and the sagittal plane are able to explain more of the variance than stiffness in a single plane alone. Although hip muscular stiffness did not result in the relationship with landing biomechanics exactly as we predicted, our findings still suggest that stiffness of the hip musculature may allow for landing biomechanics that may have the ability to lessen risk for ACL injury.

Limitations

Muscular stiffness is one of our joints' first lines of defense against injury because it is able to protect passive structures before we are able to actively compensate to a potentially injurious mechanism. When evaluating muscular stiffness in the laboratory setting, subjects are able to prepare for the perturbation by increasing their muscular activation, and therefore may appear to have greater stiffness. As muscle activity increases, more cross bridges are formed which also increases muscular stiffness (Blackburn, et al. 2004b).

Subject positioning during the stiffness assessment was also difficult to completely standardize and make the subject comfortable. As seen in figure #, subjects were positioned sidelying on the testing table with their non test leg's iliac crest at the edge of the table. We placed extra towels under their bottom leg however several subjects reported significant discomfort on their non test hip from the edge of the not allowing them to completely relax between trials opening the potential for fatigue to influence results during this assessment. Stiffness was also evaluated in the open chain and then analyzed with biomechanical results from a task in the closed chain. Future research should attempt to create an investigation that is able to evaluate hip muscular stiffness in the closed chain with an unanticipated perturbation. This may help mimic a more functional and realistic assessment of hip muscular stiffness.

In this investigation we used a double leg task primarily in the sagittal plane which may not accurately represent the physical demands of certain sports or be complex enough to evaluate the relationship of neuromuscular control and multiplanar biomechanics. Also our motion capture analysis did not include foot biomechanical data, therefore our model is not complete due to the lack of full segmental data. Future research should include a task that is multi-directional in nature and include foot biomechanical data during motion capture analysis.

Conclusions

Our findings suggest that there is a relationship between hip abductor stiffness and knee valgus motion. Muscular stiffness can be altered (Kubo, et al. 2006a) and therefore should be included in ACL injury prevention programs due to its relationship with knee valgus. Together, hip abductor and extensor muscular stiffness explained one third of the variance in knee valgus which suggests that this biomechanical risk factor for ACL injury is influenced by neuromuscular control is multiple planes. Future research is needed to further evaluate hip muscular stiffness and its ability to control ACL injury risk factors as well as its potential for intervention.

FIGURES

Figure 1: EMG electrode placement - GMax

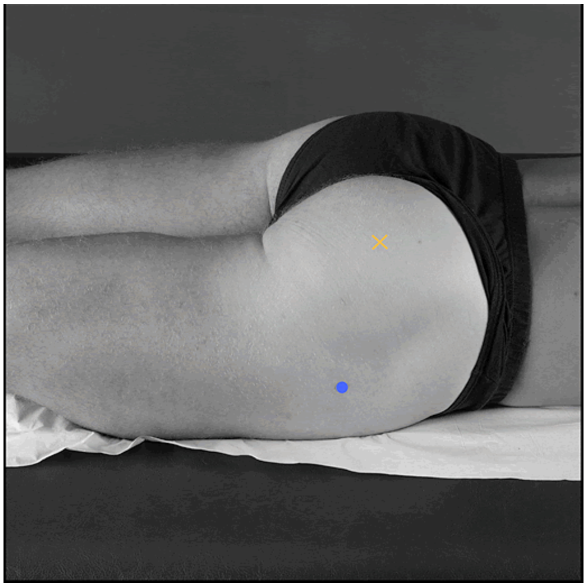


Figure 2: EMG electrode placement - GMed

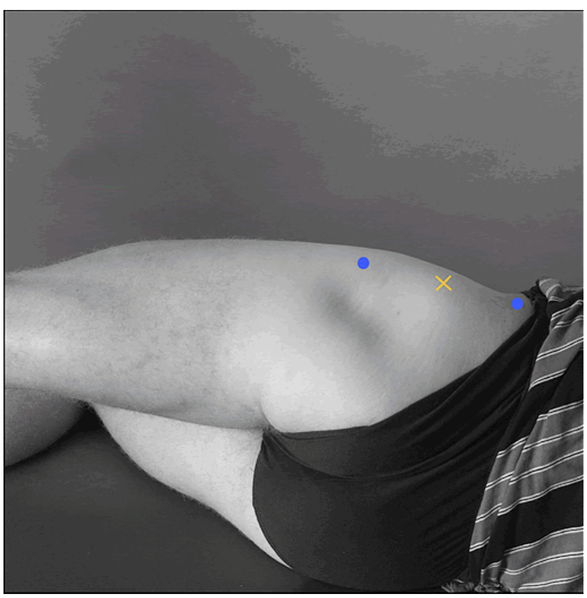


Figure 3: EMG electrode placement - MHam

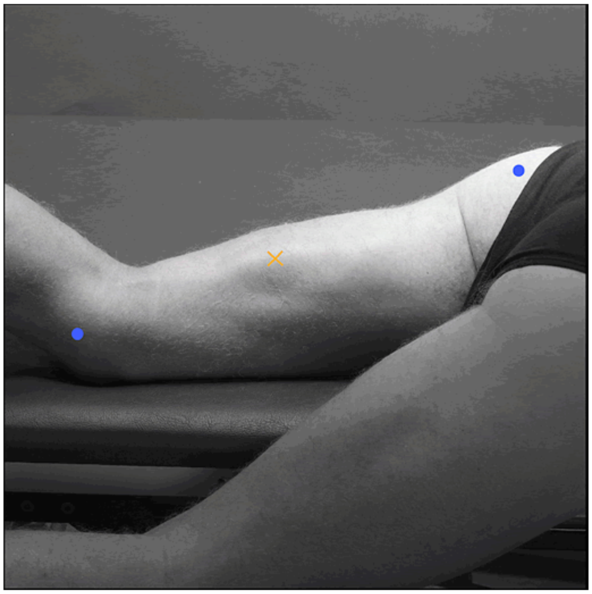


Figure 4: EMG electrode placement - LHam

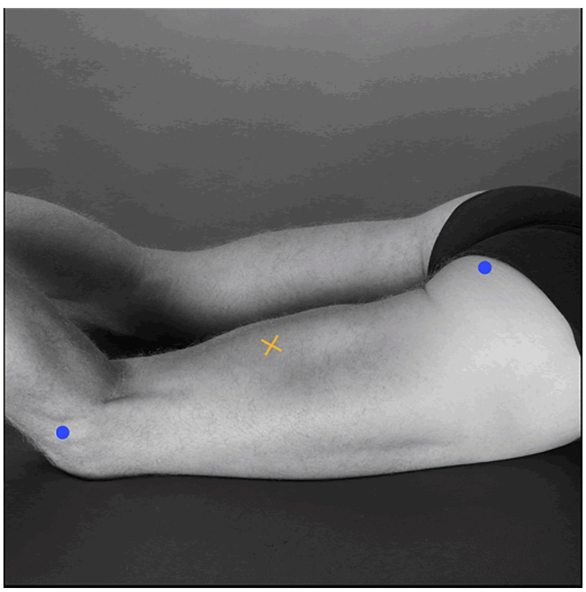


Figure 5: Stiffness assessment position - Extensor

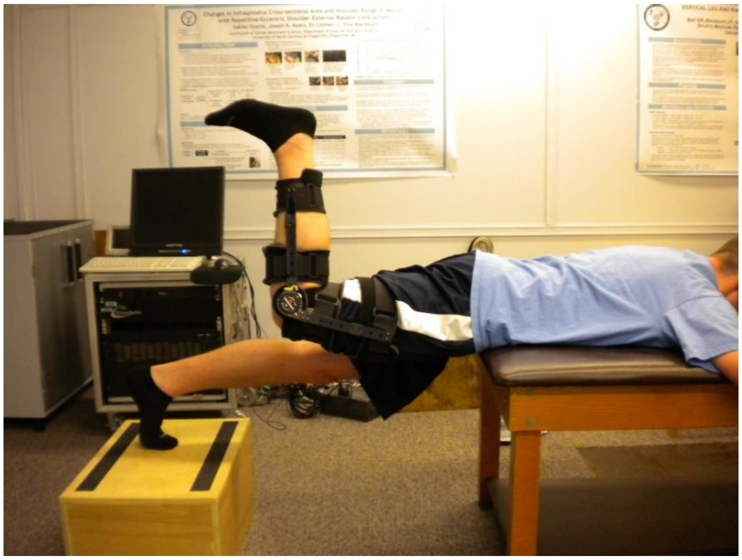


Figure 6: Stiffness assessment position - Abductor

Figure 7: Jump landing task

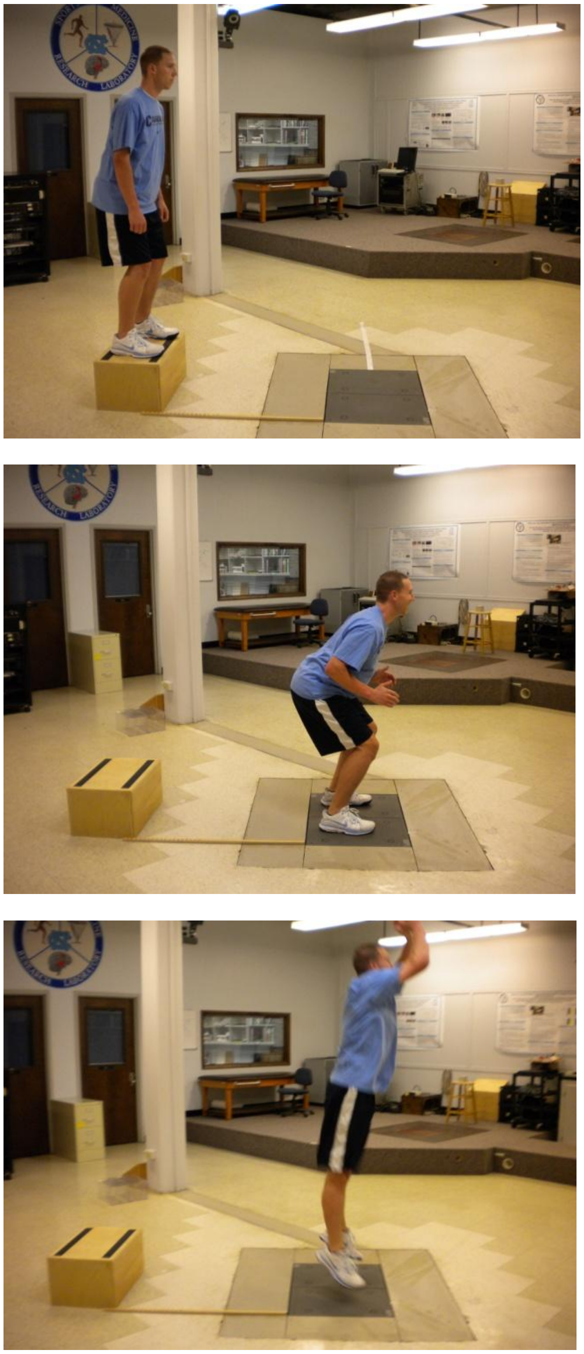


Figure 8: Abductor stiffness vs. knee valgus angle at IGC

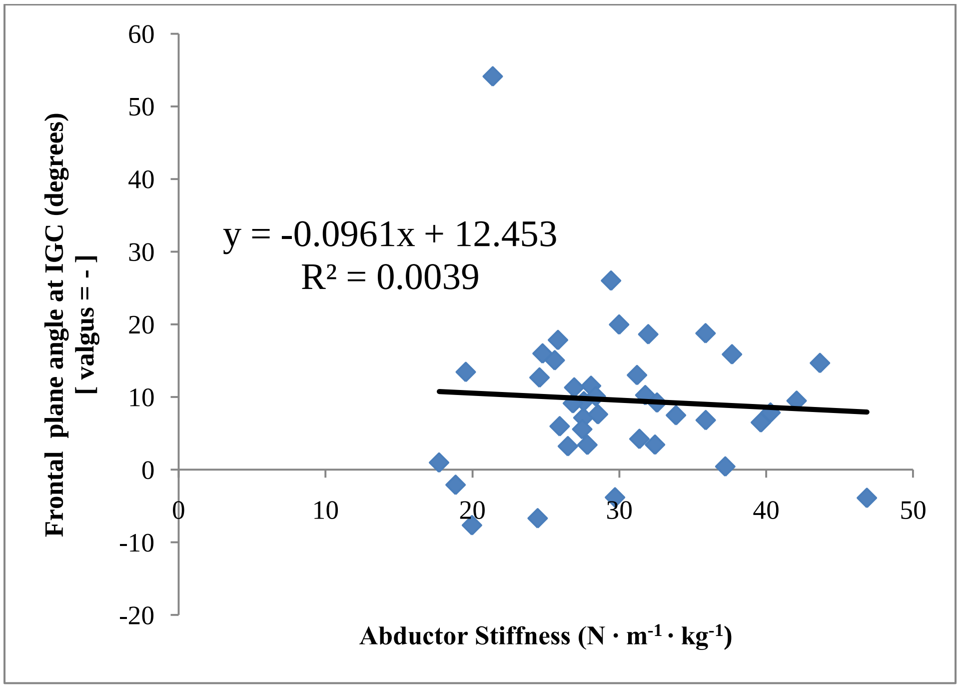


Figure 9: Abductor stiffness vs. peak knee valgus

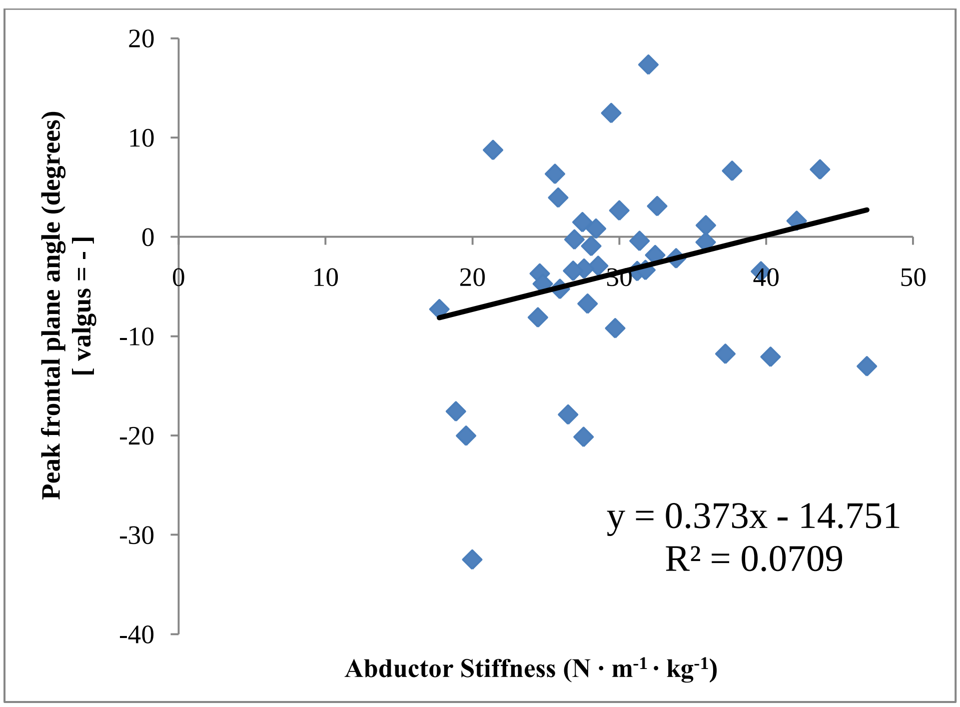


Figure 10: Abductor stiffness vs. knee valgus displacement

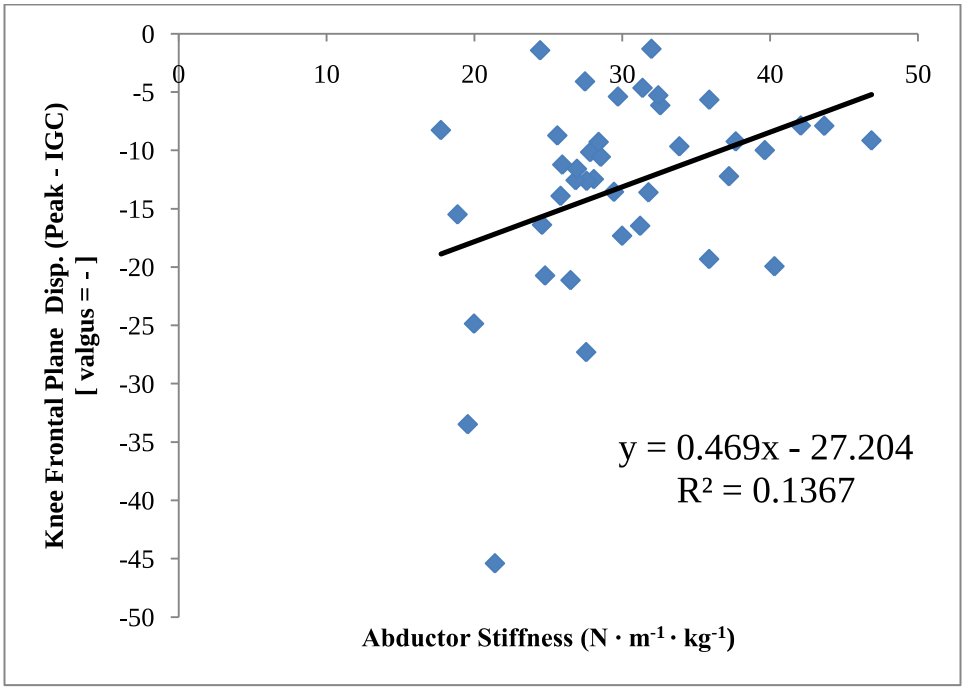


Figure 11: Abductor stiffness vs. peak internal knee varus moment

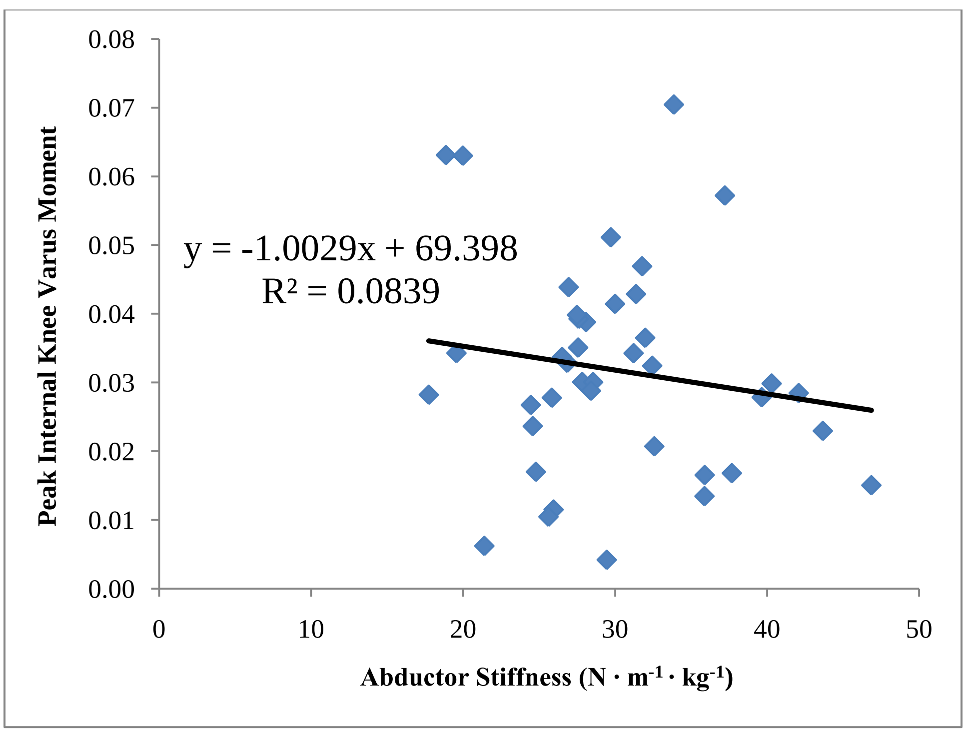


Figure 12: Abductor stiffness vs. hip adduction angle at IGC

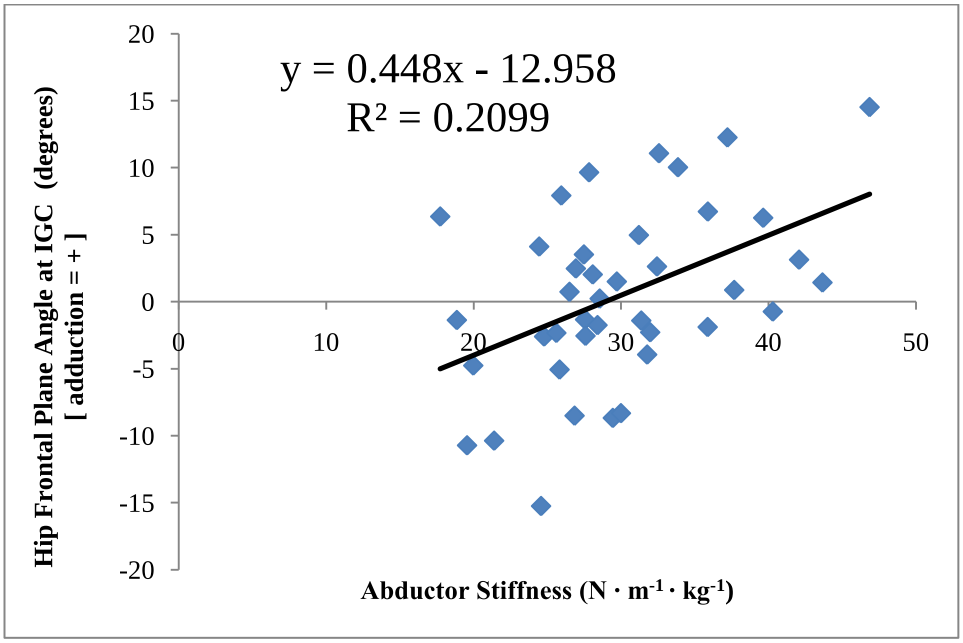


Figure 13: Abductor stiffness vs. peak hip adduction angle

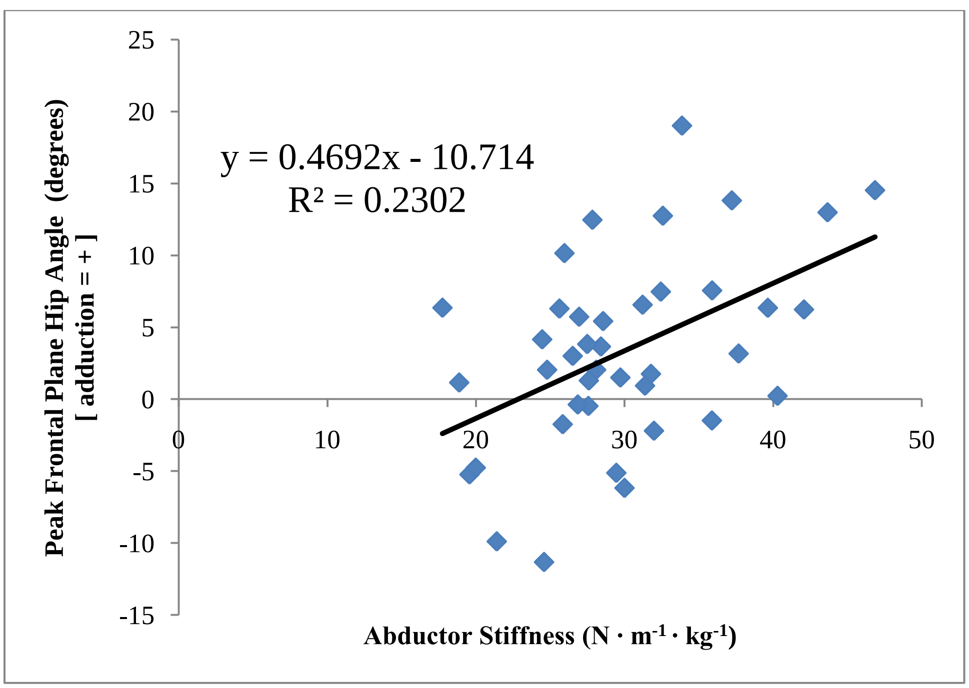


Figure 14: Abductor stiffness vs. hip adduction displacement

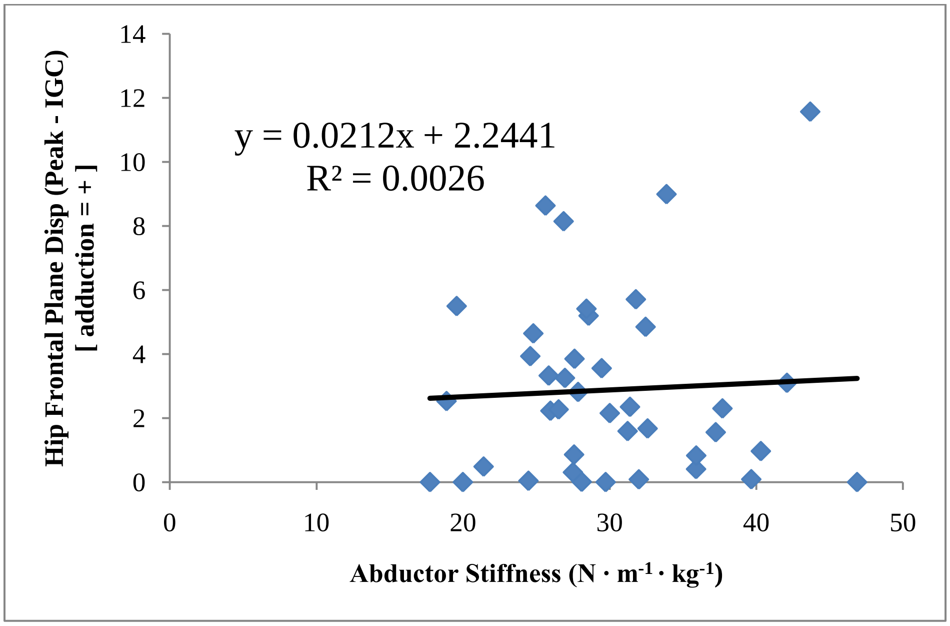


Figure 15: Abductor stiffness vs. peak hip internal abduction moment

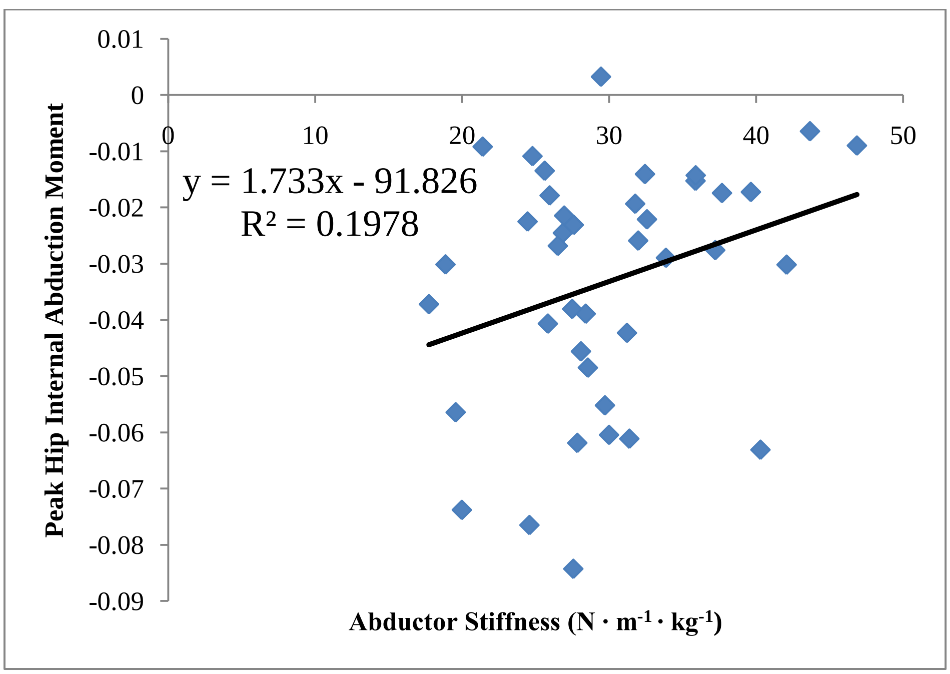


Figure 16: Extensor stiffness vs. hip flexion at IGC

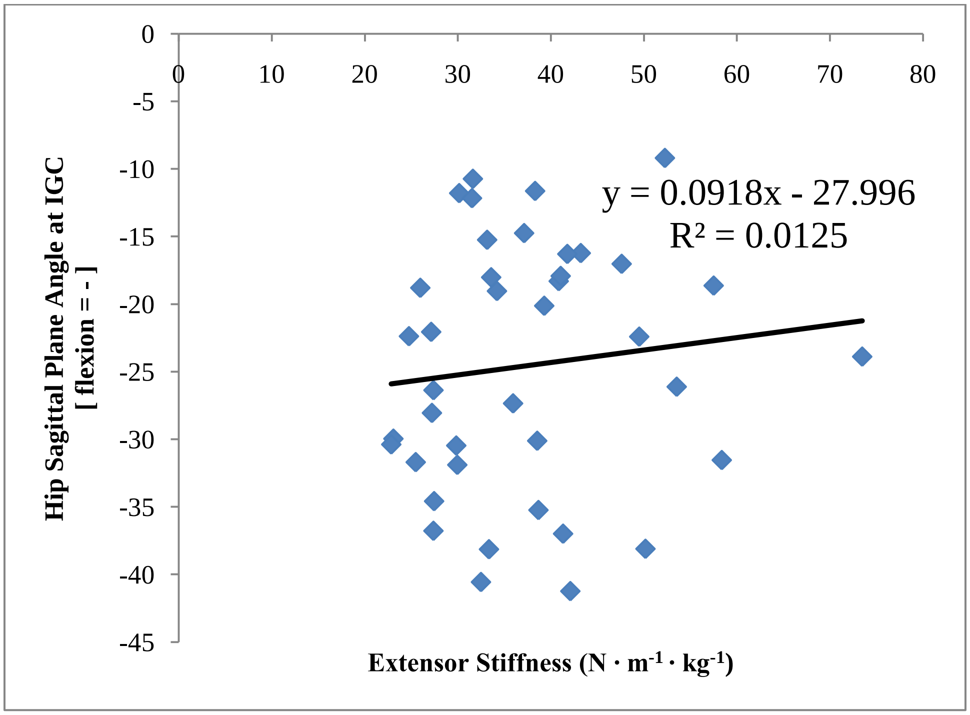


Figure 17: Extensor stiffness vs. peak hip flexion

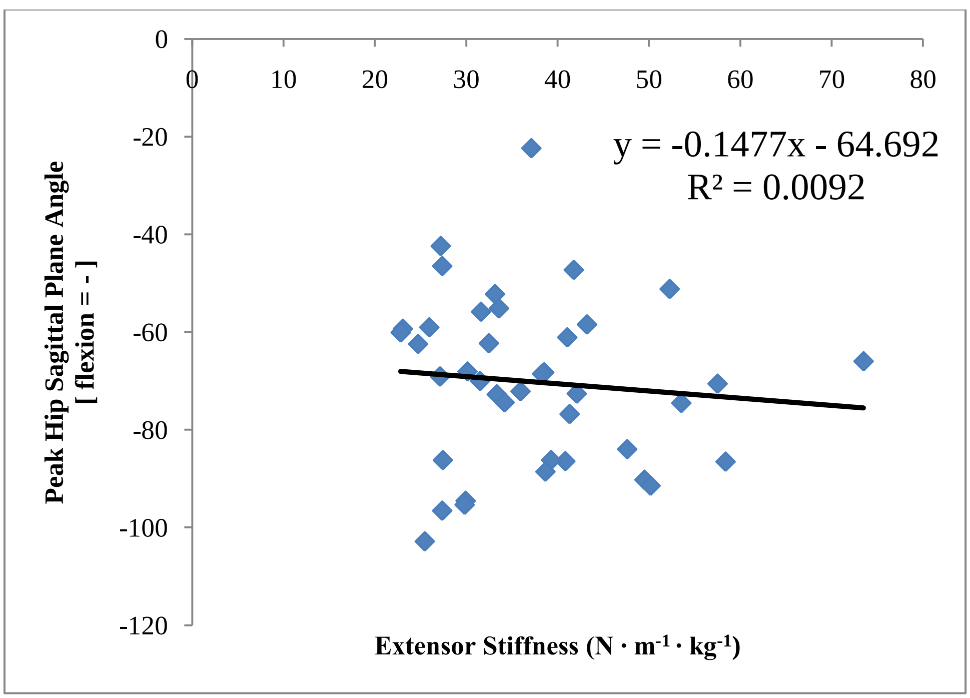


Figure 18: Extensor stiffness vs. hip flexion displacement

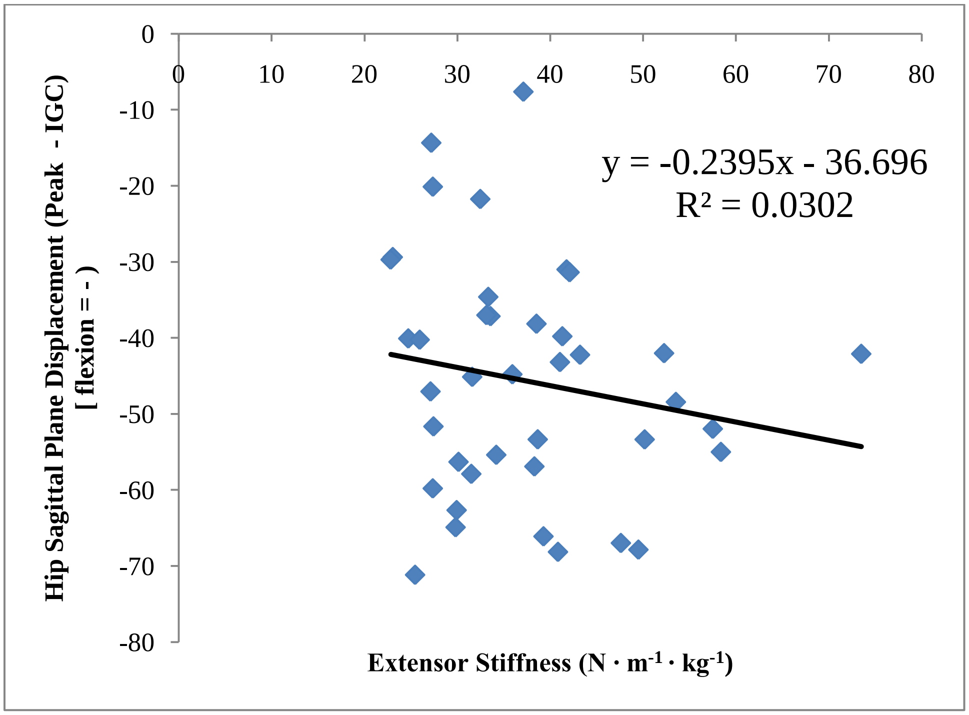


Figure 19: Extensor stiffness vs. vertical ground reaction force.

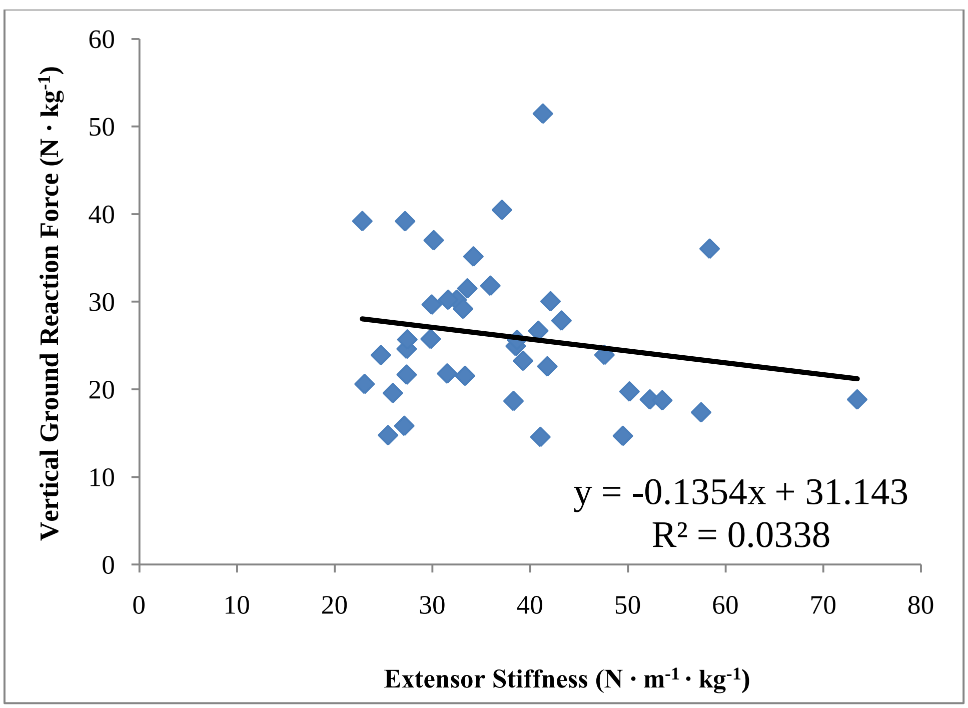


Figure 20: Extensor stiffness vs. internal rotation at IGC

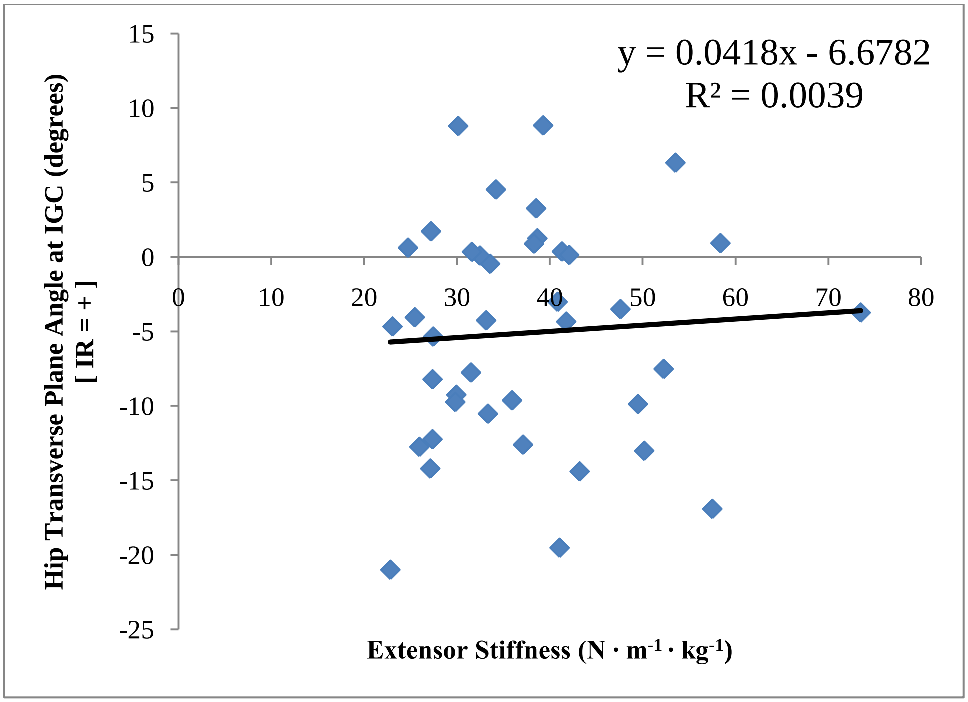


Figure 21: Extensor stiffness vs. peak hip internal rotation

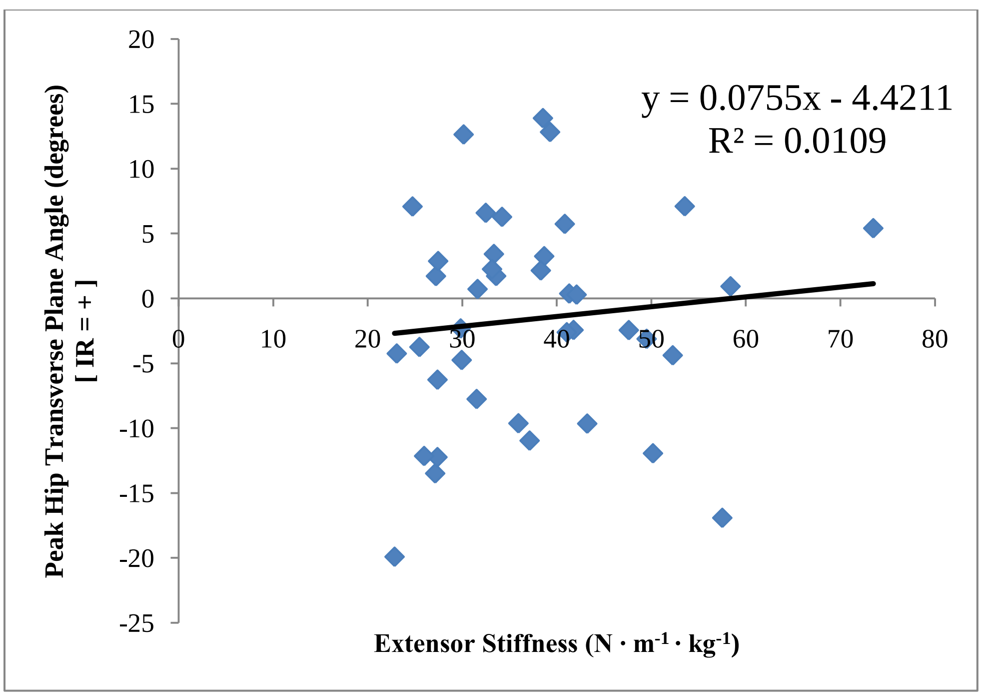
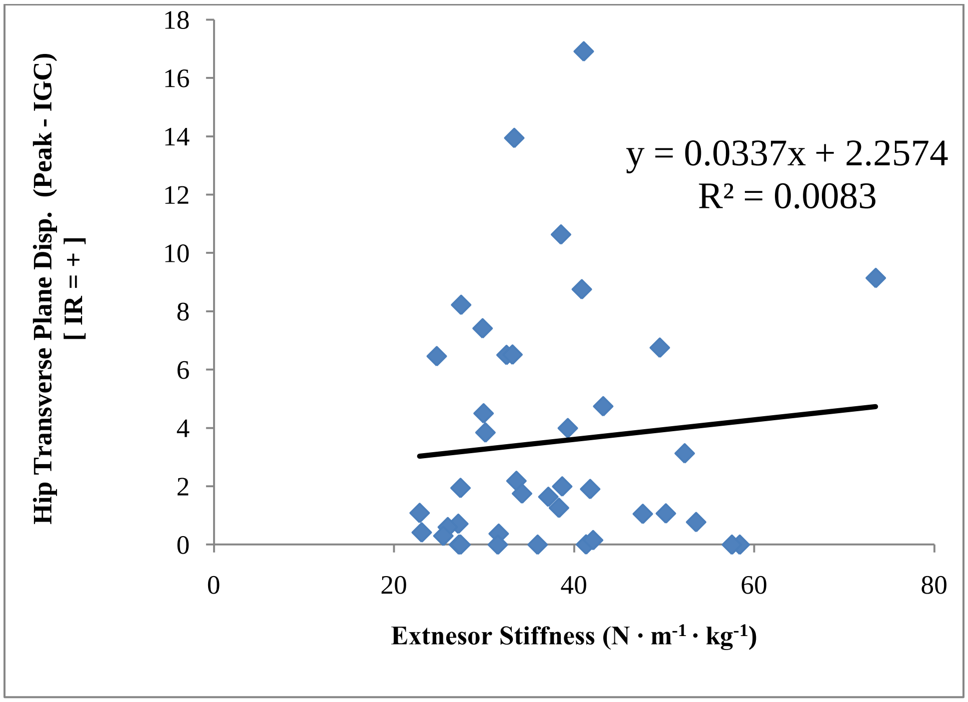


Figure 22: Extensor stiffness vs. hip internal rotation displacement



TABLES

Table #1: Research Questions and Statistical Analyses

#	Research Question	Variables	Analysis
1	Is there a significant relationship between hip abductor stiffness and the following variables during a jump landing task and side cutting task: A. knee valgus angle at initial ground contact? B. peak knee valgus angle? C. knee valgus angular displacement? D. peak internal varus moment?	Predictor: Hip abductor stiffness Criterion: A. Knee valgus angle at IC B. Peak knee valgus C. Knee valgus anglular displacement D. Peak internal knee varus moment	4 separate Pearson correlations
2	Is there a significant relationship between hip abductor stiffness and the following variables during a jump landing task and side cutting task: A. hip adduction angle at initial ground contact? B. peak hip adduction angle? C. hip adduction angular displacement? D. peak internal hip abduction moment?	Predictor: Hip abductor stiffness Criterion: A. Hip adduction angle at IC B. Peak hip adduction C. Hip adduction anglular displacement D. Peak internal hip abduction moment	4 separate Pearson correlations
3	Is there a significant relationship between hip extensor stiffness and the following variables during a jump landing task and side cutting task: A. hip flexion angle at initial ground contact B. peak hip flexion angle C. peak hip flexion angular displacement D. peak vertical ground reaction force	Predictor: Hip extensor stiffness Criterion: A. Hip flexion angle at IC B. Peak hip flexion angle C. Hip flexion angular displacement D. Peak vertical ground reaction force	4 separate Pearson correlations

#	Research Question	Variables	Analysis
4	Is there a significant relationship between hip extensor stiffness and the following variables during a jump landing task and side cutting task: A. internal rotation at initial ground contact? B. peak internal rotation angle? C. internal rotation angular displacement?	<u>Predictor:</u> Hip extensor stiffness <u>Criterion:</u> A. Internal rotation at IC B. Peak internal rotation angle C. Internal rotation angle	3 separate Pearson correlations
5	Can hip extensor stiffness and hip abductor stiffness significantly predict the following variables: A. knee valgus angle at initial ground contact? B. peak knee valgus angle? C. peak knee valgus angular displacement? D. peak vertical ground reaction force?	<u>Predictor:</u> Hip extensor stiffness Hip abductor stiffness <u>Criterion:</u> A. Knee valgus angle at IC B. Peak knee valgus angle C. Peak knee valgus angular displacement D. Peak vertical ground reaction force	4 separate multiple regressions

Table #2: Descriptive Statistics

Demographics	*Mean*	*Std. Deviation*
Height (cm)	173.425	± 9.376
Mass (kg)	71.410	± 14.517
Age	20.20	± 1.636
Kinematic Variables	*Mean*	*Std. Deviation*
Abductor Stiffness ($N \cdot m^{-1} \cdot kg^{-1}$)	29.953	± 6.786
Extensor Stiffness ($N \cdot m^{-1} \cdot kg^{-1}$)	37.468	± 11.148
Valgus at IGC	9.575	± 10.381
Peak Valgus	-3.580	± 9.510
Valgus Displacement	-13.155	± 8.610
Hip Adduction at IGC	0.462	± 6.636
Peak Hip Adduction	3.340	± 6.636
Hip Adduction Displacement	2.878	± 2.829
Hip Flexion at IGC	-24.555	± 9.170
Peak Hip Flexion	-70.225	± 17.161
Hip Flexion Displacement	-45.670	± 15.261
Hip Internal Rotation at IGC	-5.111	± 7.429
Peak Hip Internal Rotation	-1.592	± 8.052
Hip Internal Rotation Displacement	3.519	± 4.132
Kinetic Variables	*Mean*	*Std. Deviation*
Internal Knee Varus Moment	0.0318	± 0.0154
Internal Hip Abduction Moment	-0.332	± 0.0214
Peak GRFv ($N \cdot kg^{-1}$)	26.071	± 8.208

Table #3: Research Question #1 Results

Correlation between hip abductor stiffness and knee frontal plane biomechanics

	Abductor Stiffness	
Criterion Variable	r-value	p-value
Valgus at IGC	-0.063	0.350
Peak Valgus	0.266	0.048*
Valgus Displacement	0.370	0.009*
Internal Varus Moment	-0.153	0.173

*Correlation is significant at the 0.05 level (1-tailed)

Table #4: Research Question #2 Results

Correlation between abductor stiffness and hip frontal plane biomechanics

	Abductor Stiffness	
Criterion Variable	r-value	p-value
Hip Adduction at IGC	0.458	0.001*
Peak Hip Adduction	0.480	0.001*
Hip Adduction Displacement	0.051	0.378
Internal Hip Abduction moment	0.291	0.034

*Correlation is significant at the 0.05 level (1-tailed)

Table #5: Research Question #3 Results

Correlation between hip extensor stiffness and hip sagittal plane biomechanics

	Extensor Stiffness	
Criterion Variable	r-value	p-value
Hip Flexion at IGC	0.112	0.246
Peak Hip Flexion	-0.096	0.278
Hip Flexion Displacement	-0.174	0.142
Peak GRFv	-0.184	0.128

Table #6: Research Question #4 Results

Correlation between hip extensor stiffness and hip transverse plane biomechanics

	Extensor Stiffness	
Criterion Variable	r-value	p-value
Hip Internal Rotation at IGC	0.063	0.350
Peak Hip Internal Rotation	0.105	0.260
Hip Internal Rotation Displacement	0.091	0.289

Table #7: Research Question #5 Results

Multiple Regression Results

Variable	*beta*	*t*	*p-value*	R^2
Valgus at IGC				
Intercept	3.091	0.475	0.638	
Abductor Stiffness	-0.566	-2.501	0.017	0.360*
Extensor Stiffness	0.625	4.542	< 0.001	
Equation	$y = -0.556x_1 + 0.625x_2 + 3.091$			
Peak Valgus				
Intercept	-18.887	-2.755	0.009	
Abductor Stiffness	0.166	0.695	0.491	0.154*
Extensor Stiffness	0.276	1.905	0.065	
Equation	$y = 0.166x_1 + 0.276x_2 – 18.887$			
Valgus Displacement				
Intercept	-21.979	-3.887	< 0.001	
Abductor Stiffness	0.731	3.721	0.001	0.298*
Extensor Stiffness	-0.349	-2.918	0.006	
Equation	$y = 0.731x_1 – 0.349x_2 – 21.979$			
Internal Varus Moment				
Intercept	0.047	4.021	< 0.001	
Abductor Stiffness	0.000	-0.272	0.787	0.064
Extensor Stiffness	0.000	-1.273	0.211	
Equation	$y = 0.000x_1 + 0.000x_2 + 0.047$			
Peak GRFv				
Intercept	35.402	5.673	< 0.001	
Abductor Stiffness	-0.218	-1.006	0.321	0.060
Extensor Stiffness	-0.075	-0.566	0.575	
Equation	$y = -0.218x_1 – 0.075x_2 + 35.402$			

*Multiple regression significant at a 0.05 level (1-tailed)

APPENDIX

Relationship between Hip Muscle Stiffness and Biomechanical Factors Associated with ACL Injury

Tyler Cram, BS, ATC; J. Troy Blackburn, PhD, ATC;[1] Darin A. Padua, PhD, ATC; [2] Marc F. Norcross, MA, ATC[1]

[1] Neuromuscular Research Laboratory, [2] Sports Medicine Research Laboratory, University of North Carolina at Chapel Hill

Abstract

Background**:** Proper hip neuromuscular control is important for sound knee kinematics during dynamic tasks. Active muscular stiffness is believed to protect joints from injury by limiting excess joint motion. Using a proximal control theory, increased hip muscular stiffness may help protect the knee from specific movement patterns that have been shown to increase ACL injury risk.

Methods**:** Hip abductor and hip extensor stiffness was assessed in 40 physically active subjects and compared with their biomechanical data collected during a jump landing task using simple correlations. Correlations were used to evaluate relationships between hip muscular stiffness and biomechanical data collected during the dynamic task. A multiple regression was also used to evaluate the ability of hip muscular stiffness to predict movement patterns that may influence ACL injury risk.

***Findings*:** Individuals with greater hip abductor stiffness displayed lesser peak knee valgus and knee valgus displacement during a jump landing task. Greater hip abductor stiffness was also correlated with lesser hip abduction angles when compared to those with lesser hip abductor stiffness. Hip extensor stiffness was no correlated with selected variables during the jump landing task however in linear combination with hip abductor stiffness they were able to predict 36% of the variance in knee valgus angle at ground contact, 15.4% of the variance in peak knee valgus angle, and 29.8% of the variance in knee valgus displacement.

***Interpretation*:** Our findings suggest a link between hip abductor stiffness with knee valgus motion and hip adduction motion which have ACL injury implications. For the reason that muscular stiffness can be altered, it should potentially be included in ACL injury prevention programs due to its relationship with knee valgus. Together, hip abductor and extensor muscular stiffness explained one third of the variance in knee valgus which suggests that this biomechanical risk factor for ACL injury is influenced by neuromuscular control is multiple planes. Future research is needed to further evaluate hip muscular stiffness and its ability to control ACL injury risk factors as well as its potential for intervention.

Key Terms: hip muscular stiffness, ACL injury, landing biomechanics

Introduction

Noncontact anterior cruciate ligament (ACL) injury typically occurs during deceleration, lateral pivoting, or landing tasks. Ireland (1999) describes a lower extremity motion pattern known as the "position of no return" during which ACL injury commonly occurs. This pattern is characterized by adduction and internal rotation of the hip, knee valgus, and external rotation of the tibia. Hewett et al. (2005) demonstrated that peak knee valgus angle and external knee valgus moment during landing were significant prospective predictors of ACL injury risk. Additionally, participants who subsequently suffered noncontact ACL injury produced significantly greater peak vertical ground reaction forces compared to uninjured participants. These findings suggest that limiting ground reaction forces, knee valgus motion, and knee valgus moment may reduce ACL injury risk.

During closed kinematic chain activities, hip adduction results in knee valgus motion (Jacobs, et al. 2007). Knee valgus during dynamic activities have been associated with a lack of neuromuscular control of the hip (Imwalle, et al. 2009; Mclean, et al. 2005). During landing and cutting the hip abductors undergo eccentric loading in order to control hip adduction. Muscle stiffness (k) is defined as the ratio of change in force (ΔF) to change in length (ΔL) of the muscle-tendon unit ($k=\Delta F/\Delta L$) (Blackburn, et al. 2009). Given the same applied force, greater hip abductor stiffness should result in a smaller change in length, thereby limiting hip adduction and the associated knee valgus motion and moment. Chaudhari et al. (2006) demonstrated that increasing active hip joint stiffness in a frontal plane knee model decreased ACL injury threshold. However, we are unaware of any previous research which has evaluated the influence of hip muscle stiffness on landing biomechanics *in vivo*.

Landing in an erect posture characterized by a more extended knee, hip, and trunk increases vertical ground reaction forces when compared to a more flexed position (Blackburn & Padua 2009; Chappell, et al. 2007; Quatman & Hewett 2009). Conversely, landing with a more flexed trunk results in greater peak knee and hip flexion and decreases peak ground reaction forces (Blackburn & Padua 2009). In a more flexed position, the trunk's center of mass is farther from the sagittal plane hip joint axis of rotation and places a greater eccentric load on the hip extensors. Individuals with inefficient hip extensors may not be able to control trunk flexion during landing, and may compensate by landing with a more upright posture. Given the same applied force, greater hip extensor muscular stiffness should enhance the ability to control trunk flexion during landing, allowing for a more flexed landing position and decreased vertical ground reaction forces.

Greater peak vertical ground reaction forces, knee valgus angles, and knee valgus moments, and lesser peak hip flexion angles during landing have been associated with noncontact ACL injury. Greater stiffness of the hip extensors and abductors may protect the knee by limiting hip adduction and allowing for a more flexed landing posture. However, we are unaware of any research *in vivo* evaluating the influence of hip extensor and abductor stiffness on biomechanical ACL injury risk factors. Therefore the purpose of our study was to evaluate the relationships between: (1) hip abductor stiffness and knee valgus angles and moments, as well as hip adduction angles and moments; and (2) hip extensor stiffness and hip flexion angles and vertical ground reaction forces.

Methods

Subjects

A convenience sample of 40 healthy subjects (20 males and 20 females) volunteered to participate in this investigation. Subjects were recreationally active, participating in at least 30 minutes of exercise 3 times per week, and were 18-30 years of age. Potential subjects were excluded if they sustained any injury to their lower extremity in the 6 months prior to data collection that limited their activity for at least 3 consecutive days, or had a history of chronic lower extremity injury, lower extremity or spine surgery, or neurological disorder. Subjects read and signed an informed consent document approved by the university's Institutional Review Board prior to participation.

Experimental Procedures

Subjects reported to the laboratory for a single testing session. Subjects were first prepared for electromyography (EMG) electrode placement, followed by maximum voluntary isometric contractions (MVICs), muscle stiffness assessments, and a dynamic movement assessment in a counterbalanced order. All data were sampled from the dominant leg, defined as the leg they would use to kick a ball for maximum distance. Preamplified/active surface EMG electrodes (DelSys Inc., Boston, MA: interelectrode distance = 10 mm; amplification factor = 10,000 (20–450 Hz); CMMR at 60 Hz > 80 dB; input impedance > 10^{15} Ω/0.2 pF) were placed over the muscle bellies of the gluteus maximus (GMax), gluteus medius (GMed), medial hamstrings (MHam), and biceps femoris long head (LHam) parallel to the direction of action potential propagation. A reference electrode was placed on the bony prominence on the proximal antero-medial tibia. Electrodes were placed between innervation zones and distal attachments as described by Rainoldi et al.(2004). Correct electrode placements were verified using an oscilloscope during manual muscle testing (Kendall, et al. 2005).

Maximum Voluntary Isometric Contractions

Following electrode placement, all subjects completed the maximum voluntary isometric contraction (MVIC) assessments in a counterbalanced order. For hip extension, subjects were positioned prone with the anterior iliac crests (ASIS) located at the end of the table, the hip positioned in neutral, and the knee in 90° of flexion. Subjects were instructed to extend the hip with maximal effort while resistance was applied just proximal to the popliteal fossa on the posterior aspect of the thigh with a hand-held dynamometer (Lafayette instrument, Lafayette, Indiana). For hip abduction, subjects were positioned sidelying on the table on the non-dominant side with the dominant leg in full extension. Subjects were instructed to abduct the hip with maximal effort while resistance was applied just proximal to the lateral joint line of the knee with the hand-held dynamometer. Lastly, for knee flexion, subjects were positioned prone on the table with the knee in 90° of flexion and the shank perpendicular to the floor. Subjects were instructed to flex their knee for maximal effort while resistance was applied to the distal shank with the hand-held dynamometer.

Subjects completed three 5 second trials for each muscle group during which EMG amplitudes were sampled. Norcross et al. (2009) demonstrated high reliability for MVIC activation amplitudes between trials for GMax and GMed ($ICC_{2,1}$ =0.95, SEM= 49.4mV and $ICC_{2,1}$ 0.98, SEM = 50.4mV respectively). Peak forces measured during the MVIC trials using the handheld dynamometer were recorded. These results were averaged and used for loading during muscular stiffness assessment which is calculated at 30% of MVIC.

Muscle Stiffness Assessments

Muscle stiffness was calculated by modeling the hip joint as a single degree of freedom mass spring system and observing the damping effects of the hip extensors and abductors following an unanticipated perturbation (Blackburn, et al. 2004b). Five trials were completed for each muscle group with a minute of rest between trials to minimize the potential effects of fatigue. Good reliability was found for both hip extensor ($ICC_{2,1}$= 0.83, SEM = 551.7 Nm) and hip abductor stiffness ($ICC_{2,1}$= 0.74, SEM = 800.46 Nm) during pilot testing. The order of the stiffness conditions was counterbalanced.

For the hip extensor stiffness assessment, subjects were positioned prone on a table with the ASIS located at the edge of the table. The hip was maintained in neutral and the testing knee was locked in 90° of flexion in a post-operative knee brace (Bledsoe Brace Systems, Grand Prairie, TX). An accelerometer (PCB Piezotronics, Depew, NY) was placed on the lateral hinge of the knee brace and a load equaling 30% of the hip extension MVIC force was placed on the posterior aspect of the thigh proximal to the popliteal fossa using cuff weights as seen in figure 5. The thigh was then placed parallel to the floor and subjects were instructed to extend the hip isometrically to support the thigh in this position. The investigator applied a downward perturbation to the posterior thigh forcing the hip into flexion. This was done at random within 5 seconds following contraction during which accelerometer and surface EMG data were sampled.

Procedures for the hip abductor stiffness assessment were identical to those of the extensor assessment with the exception that subjects were positioned sidelying on the non-dominant side with the dominant knee locked in extension via the post-operative brace with the hip was maintained in neutral as seen in figure 6. A load equaling 30% of the hip abduction MVIC force was placed on the lateral aspect of the thigh immediately proximal to the lateral knee joint line. The test leg was positioned parallel to the floor and subjects were instructed to abduct the hip isometrically to maintain this position. The investigator then applied a downward perturbation to the lateral thigh forcing the leg into adduction. This was done at random within 5 seconds during which accelerometer and surface EMG data were sampled.

Dynamic Movement Assessment

A jump landing task was used for the dynamic movement assessment. Electromagnetic sensors (Motion Star; Ascension Technologies Inc, Burlington, VT) were secured to the pelvis, shank, and thigh of the dominant leg using double sided tape, pre-wrap, and athletic tape. The world and segment axis systems were established with the X axis defined as positive forward/anteriorly, positive Y leftward/mdeially, and positive Z upward/superiorly (Blackburn & Padua 2009). A segment-linkage model of the lower extremity was then created by digitizing the pelvis, knee, and ankle joint centers. The joint center of the knee and ankle was estimated as the midpoint between the medial and lateral malleoli and the medial and lateral femoral condyles respectively. The pelvis joint center was calculated using the bell method and digitized using the right and left ASIS (Bell, et al. 1990).

Subjects jumped from a 30 cm box to a force plate (model 4060-NC; Bertec Corp, Columbus, OH) that was located a distance equal to 50% of the participant's height in front of the box (Padua, et al. 2009). Immediately after landing, subjects completed a vertical jump for maximum height. Subjects were instructed to land on both feet, keeping the dominant foot on the force plate and to focus on gaining maximum vertical height after landing. Subjects were allowed as many practice trials as needed to feel comfortable with the task. A total of 5 successful trials were completed for each subject with a minute of rest between trials to reduce effect of fatigue. For a successful trial, subjects had to jump off the box with both feet, land with their dominant foot completely on the force plate and their non-dominant foot completely off force plate. If these criteria are not met the trial was repeated.

Data Sampling and Reduction

Electromagnetic sensor coordinate data were sampled at 100 Hz, while force plate, accelerometer, and EMG data were sampled at 1,000 Hz. Kinematic data were time synchronized with EMG, kinetic, and accelerometer data and re-sampled at 1,000 Hz via linear interpolation. All data were captured and synchronized using The MotionMonitor software (Innovative Sports Training: Chicago, IL).

EMG data were corrected for DC bias; bandpass (20-350Hz) and notch (59.5-60.5Hz) filtered; then smoothed using a 25 ms root mean square sliding window function (Norcross, et al. 2009). All EMG data were normalized to mean activation amplitude during the middle three seconds of the respective GMax/GMed/MHam/LHam MVIC trials.

Accelerometer data were low pass filtered at 10Hz with a fourth order zero phase lag Butterworth filter. The first two oscillatory peaks in the tangential leg acceleration profile were identified and used to calculate the damped frequency of oscillation $\frac{1}{t_2 - t_1}$. This value was then be used to calculate the linear stiffness (Blackburn, et al. 2004b) using the equation $k = 4\pi^2 m f^2$, where k is linear stiffness, m is the summed mass of the system (thigh, shank, foot, brace and the applied load), and f is the damped frequency of oscillation. Because stiffness increases as a function of mass (Blackburn, et al. 2009), these data were normalized to body mass prior to statistical analyses.

Kinematic data were lowpass filtered at 10 Hz with a 4^{th} order zero phase lag Butterworth filter. Euler angles were used for all kinematic calculations and were calculated in an YX'Z" flexion/extension, abduction/adduction, and internal/external rotation sequence. Knee angles were calculated as the shank relative to the thigh and hip angles were calculated as the thigh relative to pelvis. Kinetic data were low-pass filtered at 50Hz with a fourth order zero phase lag Butterworth filter and combined with kinematics via an inverse dynamics solution was to calculate internal frontal plane knee moments.

Peak knee valgus angle, internal knee varus moment, hip flexion angle, and vertical ground reaction force were calculated during the loading phase of the jump landing task. Frontal plane knee and sagittal plane hip angles at initial contact and total displacement during the loading phase were also calculated. The loading phase was defined as the interval from initial ground contact (vertical ground reaction force > 10 N) to peak knee flexion. Ground reaction forces were normalized to body weight and moments were normalized to the product of the subject's mass and height.

All statistical analyses will be performed using SPSS version 18.0. Simple Pearson correlations will be used to assess muscular stiffness and biomechanical relationships during a jump landing task. We will analyze the relationship of hip abductor stiffness with knee valgus, hip adduction, and hip internal abductor moment.

We will also analyze the relationship between hip extensor stiffness with hip flexion, hip internal rotation and vertical ground reaction force. A prediction model will be assessed using multiple regression analysis with an order of entry beginning with the highest simple correlation. Statistical significance was established a priori as $\alpha \leq 0.05$.

Results

While the relationship between hip abductor stiffness and the knee valgus angle at initial ground contact was non-significant ($r = -0.063$, $P = 0.350$), hip abductor stiffness was significantly correlated with peak valgus angles ($r = 0.266$, $P = 0.048$) as well as valgus angular displacement ($r = 0.370$, $P = 0.009$). Knee valgus was given a negative angular convention, therefore these data indicate that subjects with greater hip abductor stiffness displayed lesser peak knee valgus angles and valgus displacement. The relationship between hip abductor stiffness and the peak internal knee varus moment was non-significant ($r = -0.153$, $P = 0.173$). Hip abductor stiffness was significantly correlated with hip adduction angles at initial ground contact ($r = 0.458$, $P = 0.001$), peak hip adduction angle ($r = 0.480$, $P = 0.001$), and peak internal hip abduction moment ($r = 0.291$, $P = 0.034$). Hip adduction was given a positive angular convention, therefore these results indicate that subjects with greater hip abductor stiffness landed with greater hip adduction at initial ground contact, and displayed greater peak hip adduction angles and lesser internal hip abduction moment compared to those with less stiff/more compliant hip abductors. The relationship between abductor stiffness and hip adduction angular displacement was not significant ($r = 0.051$, $P = 0.378$).

Hip extensor stiffness was not significantly correlated with hip flexion at initial ground contact ($r = 0.112$, $P = 0.246$), peak hip flexion ($r = -0.096$, $P = 0.278$), hip flexion angular displacement (r = -0.174, P = 0.142), or peak vertical ground reaction force (r = -0.184, P = 0.128). Additionally, hip extensor stiffness was not significantly related to hip internal rotation at initial ground contact (r = 0.063, P = 0.350), peak hip internal rotation (r = 0.105, P = 0.260), or hip internal rotation angular displacement (r = 0.091, P = 0.289).

Using multiple regression analysis, we found that the linear combination of hip abductor stiffness and extensor stiffness was significantly related to knee valgus angle at initial ground contact (R^2 = .360, P = < 0.001), peak knee valgus angle (R^2 = 0.154, P = 0.045), and knee valgus displacement (R^2 = 0.298, P = 0.001). The linear combination of abductor stiffness and extensor stiffness was not correlated with peak internal knee varus moment (R^2 = 0.062, P = 0.292) or vertical ground reaction force (R^2 = 0.060, P = 0.321).

Discussion

The primary findings of this study showed that individuals with greater hip abductor stiffness displayed less knee valgus motion during a dynamic movement task. Also, individuals with greater hip abductor stiffness displayed greater peak hip adduction during the dynamic task. Hip extensor stiffness did not significantly correlate with any of the selected biomechanical variables. While in isolation, the predicative ability for abductor stiffness and valgus at initial contact, peak valgus, and valgus displacement was non-significant, 7%, and 14% respectively. However, when hip abductor stiffness is combined linearly with hip extensor stiffness, its predictive ability increased for valgus at intitial contact, peak valgus, and valgus displacement to 36%, 15%, and 30% respectively. These results suggest that hip muscular stiffness has the ability to influence landing biomechanics related to ACL injury risk.

To our knowledge this is the first investigation to evaluate relationships between hip muscular stiffness and landing biomechanics related to ACL injury. Therefore, comparison to previous literature is limited. However, our findings are consistent with previous literature supporting a relationship between hip neuromuscular control and biomechanical factors that are linked with knee injury (Carcia, et al. 2005; Gehring, et al. 2009; Jacobs, et al. 2007).

Fatigued muscles perform in a similar fashion as weak muscles due to altered force production (Millet & Lepers 2004). There have been several studies investigating the effect of hip abductor fatigue on landing mechanics to evaluate their influence on frontal plane ACL injury risk factors. Jacobs (2007) and Gehring (2009) evaluated gender differences of knee biomechanics following a hip abductor fatigue protocol. Both studies found that when compared to males females exhibited greater peak knee valgus during the post-fatigue landing task potentially increasing their risk for ACL injury. Using a similar study design, Carcia (2005) also investigated the effect of hip abductor fatigue on landing biomechanics. Contrary to Jacobs and Gehring and similar to our investigation, Carcia found a relationship between knee valgus and hip abductor neuromuscular control regardless of gender. They found that subjects landed with more knee valgus at initial ground contact following fatigue protocol. However, they did not find a relationship between peak valgus or valgus displacement which may have more influence on ACL injury compared to valgus angles at initial ground contact.

We confirmed our primary hypothesis that stiffer hip abductors are associated with lesser knee valgus motion. The rationale behind our hypothesis was that greater hip abductor stiffness provides greater resistance to lengthening, thus limiting hip adduction motion distally. Since hip adduction contributes to dynamic knee valgus in the frontal plane, limiting this motion will also limit the amount of valgus the knee undergoes during dynamic activities. However, we found that precisely the opposite occurred: individuals with greater hip abductor stiffness landed in a less abducted position. The explanation for this unexpected finding is not completely clear. Landing in a more adducted position may be a protective landing strategy to mechanically assist the hip abductors to control knee valgus and knee valgus displacement. Although, this is most likely not the case because the hip abductors have been shown to exhibit linear increases in strength as they move from an abducted position through neutral and into a slightly adducted position (Ryser, et al. 1988)

If individuals that have decreased hip abductor stiffness display greater knee valgus displacement, they must land in a position that will allow them to undergo enough knee valgus without causing injury. One way they may accomplish this is by landing in a more hip abducted position to allow for total knee valgus motion. This may represent a phenomenon already described in the literature know as ligament dominance which occurs when individuals allow knee ligaments rather than lower extremity musculature to absorb ground reaction forces. (Myer, et al. 2004) Ligament dominant landing preference is typically characterized by increased medial knee displacement as well as high ground reaction forces (Ford, et al. 2003). However, we did not find any relationship between hip neuromuscular control and increased ground reaction forces.

In order to better understand our results we organized the data into tertiles based on hip abductor stiffness and ran secondary analysis. We then used descriptive statistics and independent samples t-tests to analyze biomechanical data of the high stiffness group

(T1) and the low stiffness group (T3). Results from this secondary analysis were able to show both the high stiffness group and the low stiffness group went through the same amount of hip adduction displacement. Although we did not include foot positioning in our motion capture model, it is a possibility that individuals with lesser hip abduction stiffness landed with their feet further apart. If those who displayed lesser hip abduction stiffness truly landed in a wider stance it would explain why they displayed more hip abduction at initial ground contact and displayed lesser peak hip adduction angles. Also with their feet further from the midline when landing, they would also exhibit greater peak knee valgus and knee valgus displacement for the same amount of hip adduction displacement.

It may also be plausible that subjects with lesser hip abductor stiffness activated their hip abductors more in preparation for landing resulting in a more abduction hip at initial ground contact. To further evaluate this possible relationship we ran simple correlations between gluteus medius muscle activation during the jump landing task and hip adduction angles at IGC, peak adduction angle, and hip adduction displacement. Our findings from this analysis did not reveal any significant relationships. Future research needs to investigate other potential reasons to explain why those with increased muscular hip stiffness land in a more hip adducted position than those with less hip abductor stiffness.

Hip extensor stiffness was not correlated with peak hip flexion angle, internal rotation angle, or vertical ground reaction force, factors which have been previously associated with ACL injury risk (Blackburn & Padua 2009; Hewett, et al. 2005). For the purposes of our investigation we viewed the gluteus maximus as the primary hip extensor. The gluteus maximus assists in external rotation, but it is not the only hip external rotator. For this reason its stiffness alone may not adequately represent hip external rotator stiffness which could explain the lack of correlation with peak hip internal rotation during the task.

While the jump landing task may have been difficult enough to show neuromuscular deficits in the frontal plane, this task may not been complex enough to replicate realistic functional demands to reveal any relationships with hip extensor stiffness and associated injury risk factors. Blackburn (2009) was able to show that increased trunk flexion during from a box height of 60cm was able to decrease vertical ground reaction forces. Since vertical ground reaction forces during jump landing tasks increase as box height increases (Yeow, et al. 2009) we may have needed to use a higher box greater than 30cm in order to find differences in hip extensor stiffness particularly in the sagittal plane. Nevertheless, we chose this task for our investigation because it has been used previously (Hewett, et al. 2005) to evaluate and predict ACL injury risk.

Although hip extensor stiffness was not correlated with landing biomechanics in isolation, the linear combination of hip abductor and extensor stiffness predicted 36% of the variance in knee valgus angle at ground contact, 15.4% of the variance in peak knee valgus angle, and 29.8% of the variance in knee valgus displacement. While on the other hand, hip abductor stiffness was only able to significantly predict 7% of peak knee valgus and 14% of knee valgus displacement. Dynamic knee valgus is a multiplanar motion which consists of femoral adduction, femoral and tibial rotation. Since dynamic knee valgus occurs in multiple planes, it is understandable why the combination of stiffness in the frontal plane and the sagittal plane are able to explain more of the variance than stiffness in a single plane alone. Although hip muscular stiffness did not result in the relationship with landing biomechanics exactly as we predicted, our findings still suggest that stiffness of the hip musculature may allow for landing biomechanics that may have the ability to lessen risk for ACL injury.

Limitations

Muscular stiffness is one of our joints' first lines of defense against injury because it is able to protect passive structures before we are able to actively compensate to a potentially injurious mechanism. When evaluating muscular stiffness in the laboratory setting, subjects are able to prepare for the perturbation by increasing their muscular activation, and therefore may appear to have greater stiffness. As muscle activity increases, more cross bridges are formed which also increases muscular stiffness (Blackburn, et al. 2004b).

Subject positioning during the stiffness assessment was also difficult to completely standardize and make the subject comfortable. As seen in figure #, subjects were positioned sidelying on the testing table with their non test leg's iliac crest at the edge of the table. We placed extra towels under their bottom leg however several subjects reported significant discomfort on their non test hip from the edge of the not allowing them to completely relax between trials opening the potential for fatigue to influence results during this assessment. Stiffness was also evaluated in the open chain and then analyzed with biomechanical results from a task in the closed chain. Future research should attempt to create an investigation that is able to evaluate hip muscular stiffness in the closed chain with an unanticipated perturbation. This may help mimic a more functional and realistic assessment of hip muscular stiffness.

In this investigation we used a double leg task primarily in the sagittal plane which may not accurately represent the physical demands of certain sports or be complex enough to evaluate the relationship of neuromuscular control and multiplanar biomechanics. Also our motion capture analysis did not include foot biomechanical data, therefore our model is not complete due to the lack of full segmental data. Future research should include a task that is multi-directional in nature and include foot biomechanical data during motion capture analysis.

Conclusions

Our findings suggest that there is a relationship between hip abductor stiffness and knee valgus motion. Muscular stiffness can be altered (Kubo, et al. 2006a) and therefore should be included in ACL injury prevention programs due to its relationship with knee valgus. Together, hip abductor and extensor muscular stiffness explained one third of the variance in knee valgus which suggests that this biomechanical risk factor for ACL injury is influenced by neuromuscular control is multiple planes. Future research is needed to further evaluate hip muscular stiffness and its ability to control ACL injury risk factors as well as its potential for intervention.

REFERENCES

Amis, A., Gupte, C., Bull, A., & Edwards, A. (2006). Anatomy of the posterior cruciate ligament and the meniscofemoral ligaments. *Knee Surgery, Sports Traumatology, Arthroscopy, 14*(3), 257-263.

Bell, A. L., Pedersen, D. R., & Brand, R. A. (1990). A comparison of the accuracy of several hip center location prediction methods. *J Biomech, 23*(6), 617-621.

Bell, D. R., Myrick, M. P., Blackburn, J. T., Shultz, S. J., Guskiewicz, K. M., & Padua, D. A. (2009). The effect of menstrual-cycle phase on hamstring extensibility and muscle stiffness. *J Sport Rehabil, 18*(4), 553-563.

Besier, T. F., Lloyd, D. G., Cochrane, J. L., & Ackland, T. R. (2001). External loading of the knee joint during running and cutting maneuvers. *Medicine & Science in Sports & Exercise, 33*(7), 1168-1175.

Blackburn, J. T., Bell, D. R., Norcross, M. F., Hudson, J. D., & Kimsey, M. H. (2009). Sex comparison of hamstring structural and material properties. *Clin Biomech (Bristol, Avon), 24*(1), 65-70.

Blackburn, J. T., & Padua, D. A. (2008). Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clinical Biomechanics, 23*(3), 313-319.

Blackburn, J. T., & Padua, D. A. (2009). Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train, 44*(2), 174-179.

Blackburn, J. T., Padua, D. A., Riemann, B. L., & Guskiewicz, K. M. (2004a). The relationships between active extensibility, and passive and active stiffness of the knee flexors. *J Electromyogr Kinesiol, 14*(6), 683-691.

Blackburn, J. T., Riemann, B. L., Padua, D. A., & Guskiewicz, K. M. (2004b). Sex comparison of extensibility, passive, and active stiffness of the knee flexors. *Clin Biomech (Bristol, Avon), 19*(1), 36-43.

Boden, B. P., Dean, S. G., Feagin, J. A., & Garrett, W. E. (2000a). Mechanisms of Anterior Cruciate Ligament Injury. *Orthopedics, 23*(6), 573-578.

Boden, B. P., Griffin, L. Y., & Garrett, W. E., Jr. (2000b). Etiology and Prevention of Noncontact ACL Injury. *Phys Sportsmed, 28*(4), 53-60.

Carcia, C., Eggen, J., & Shultz, S. (2005). Hip-abductor fatigue, frontal-plane landing angle, and excursion during a drop jump. *Journal of Sport Rehabilitation, 14*(4), 321-331.

Chappell, J. D., Creighton, R. A., Giuliani, C., Yu, B., & Garrett, W. E. (2007). Kinematics and electromyography of landing preparation in vertical stop-jump: risks for noncontact anterior cruciate ligament injury.(Clinical report). *The American Journal of Sports Medicine, 35*(2), 235(237).

Chappell, J. D., Yu, B., Kirkendall, D. T., & Garrett, W. E. (2002). A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med, 30*(2), 261-267.

Chaudhari, A. M., & Andriacchi, T. P. (2006). The mechanical consequences of dynamic frontal plane limb alignment for non-contact ACL injury. *Journal of Biomechanics, 39*(2), 330-338.

Childs, S. G. (2002). Pathogenesis of anterior cruciate ligament injury. *Orthop Nurs, 21*(4), 35-40.

Duthon, V. B., Barea, C., Abrassart, S., Fasel, J. H., Fritschy, D., & Menetrey, J. (2006). Anatomy of the anterior cruciate ligament. *Knee Surg Sports Traumatol Arthrosc, 14*(3), 204-213.

Ford, K. R., Myer, G. D., & Hewett, T. E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Med Sci Sports Exerc, 35*(10), 1745-1750.

Gagnon, D., & Gagnon, M. (1992). The influence of dynamic factors on triaxial net muscular moments at the L5/S1 joint during asymmetrical lifting and lowering. *J Biomech, 25*(8), 891-901.

Gehring, D., Melnyk, M., & Gollhofer, A. (2009). Gender and fatigue have influence on knee joint control strategies during landing. *Clinical Biomechanics, 24*(1), 82-87.

Gelber, A. C., Hochberg, M. C., Mead, L. A., Wang, N.-Y., Wigley, F. M., & Klag, M. J. (2000). Joint Injury in Young Adults and Risk for Subsequent Knee and Hip Osteoarthritis. *Annals of Internal Medicine, 133*(5), 321-328.

Griffin, L. Y., Agel, J., Albohm, M. J., Arendt, E. A., Dick, R. W., Garrett, W. E., et al. (2000). Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *J Am Acad Orthop Surg, 8*(3), 141-150.

Hanson, A. M., Padua, D. A., Troy Blackburn, J., Prentice, W. E., & Hirth, C. J. (2008). Muscle activation during side-step cutting maneuvers in male and female soccer athletes. *J Athl Train, 43*(2), 133-143.

Hewett, T., Lindenfeld, T. N., Riccobene, J. V., & Noyes, F. R. (1999). The Effect of Neuromuscular Training on the Incidence of Knee Injury in Female Athletes. *The American Journal of Sports Medicine, 27*(6), 699-706.

Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Jr., Colosimo, A. J., McLean, S. G., et al. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med, 33*(4), 492-501.

Hewett, T. E., Torg, J. S., & Boden, B. P. (2009). Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *Br J Sports Med, 43*(6), 417-422.

Imwalle, L. E., Myer, G. D., Ford, K. R., & Hewett, T. E. (2009). Relationship between hip and knee kinematics in athletic women during cutting maneuvers: a possible link to noncontact anterior cruciate ligament injury and prevention. *J Strength Cond Res, 23*(8), 2223-2230.

Ireland, M. L. (1999). Anterior Cruciate Ligament Injury in Female Athletes: Epidemiology. *J Athl Train, 34*(2), 150-154.

Jacobs, C. A., Uhl, T. L., Mattacola, C. G., Shapiro, R., & Rayens, W. S. (2007). Hip abductor function and lower extremity landing kinematics: sex differences. *J Athl Train, 42*(1), 76-83.

Kendall, F., McCreary, E., Provance, P., Rodgers, M., & Romani, W. (Eds.). (2005). *Muscles - Testing and Function* (5th ed.). Philadelphia: Lippincott Williams & Wilkins.

Kernozek, T. W., Torry, M. R., & Iwasaki, M. (2008). Gender Differences in Lower Extremity Landing Mechanics Caused by Neuromuscular Fatigue. *The American Journal of Sports Medicine, 36*(3), 554-565.

Kubo, K., Ohgo, K., Takeishi, R., Yoshinaga, K., Tsunoda, N., Kanehisa, H., et al. (2006a). Effects of isometric training at different knee angles on the muscle-tendon complex in vivo. *Scand J Med Sci Sports, 16*(3), 159-167.

Kubo, K., Yata, H., Kanehisa, H., & Fukunaga, T. (2006b). Effects of isometric squat training on the tendon stiffness and jump performance. *Eur J Appl Physiol, 96*(3), 305-314.

Lohmander, L. S., Englund, P. M., Dahl, L. L., & Roos, E. M. (2007). The long-term consequence of anterior cruciate ligament and meniscus injuries: osteoarthritis. *Am J Sports Med, 35*(10), 1756-1769.

Magee, D. (Ed.). (2006). *Orthopedic Physcial Assessment* (4th ed.). St. Louis: Saunders.

McLean, S. G., Huang, X., Su, A., & Van Den Bogert, A. J. (2004). Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clin Biomech (Bristol, Avon), 19*(8), 828-838.

McLean, S. G., Huang, X., & van den Bogert, A. J. (2005). Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: Implications for ACL injury. *Clinical Biomechanics, 20*(8), 863-870.

Meyers, M. C., & Barnhill, B. S. (2004). Incidence, Causes, and Severity of High School Football Injuries on FieldTurf Versus Natural Grass. *The American Journal of Sports Medicine, 32*(7), 1626-1638.

Millet, G. Y., & Lepers, R. (2004). Alterations of neuromuscular function after prolonged running, cycling and skiing exercises. *Sports Med, 34*(2), 105-116.

Moore, K., & AM, A. (Eds.). (2007). *Essential Clinical Anatomy* (3rd ed.). Philadelphia: Lippincott Williams & Wilkins.

Myer, G. D., Ford, K. R., & Hewett, T. E. (2004). Rationale and Clinical Techniques for Anterior Cruciate Ligament Injury Prevention Among Female Athletes. *Journal of Athletic Training, 39*(4), 352-364.

Norcross, M. F., Troy Blackburn, J., & Goerger, B. M. (2009). Reliability and interpretation of single leg stance and maximum voluntary isometric contraction methods of electromyography normalization. *J Electromyogr Kinesiol*.

Nyland, J., Kuzemchek, S., Parks, M., & Caborn, D. N. M. (2004). Femoral anteversion influences vastus medialis and gluteus medius EMG amplitude: composite hip abductor EMG amplitude ratios during isometric combined hip abduction-external rotation. *Journal of Electromyography and Kinesiology, 14*(2), 255-261.

Padua, D. A., & Blackburn, J. T. (2003). Research digest. Muscle stiffness and biomechanical stability. *Athletic Therapy Today, 8*(6), 45-47.

Padua, D. A., Marshall, S. W., Boling, M. C., Thigpen, C. A., Garrett, W. E., Jr., & Beutler, A. I. (2009). The Landing Error Scoring System (LESS) Is a valid and reliable clinical assessment tool of jump-landing biomechanics: The JUMP-ACL study. *Am J Sports Med, 37*(10), 1996-2002.

Palmieri-Smith, R. M., & Thomas, A. C. (2009). A neuromuscular mechanism of posttraumatic osteoarthritis associated with ACL injury. *Exerc Sport Sci Rev, 37*(3), 147-153.

Quatman, C. E., & Hewett, T. E. (2009). The anterior cruciate ligament injury controversy: is 'valgus collapse' a sex-specific mechanism? *British Journal of Sports Medicine, 43*(5), 328-335.

Rainoldi, A., Melchiorri, G., & Caruso, I. (2004). A method for positioning electrodes during surface EMG recordings in lower limb muscles. *Journal of Neuroscience Methods, 134*(1), 37-43.

Ryser, D. K., Erickson, R. P., & Cahalan, T. (1988). Isometric and isokinetic hip abductor strength in persons with above-knee amputations. *Arch Phys Med Rehabil, 69*(10), 840-845.

Shultz, S. J., Kirk, S. E., Johnson, M. L., Sander, T. C., & Perrin, D. H. (2004). Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Med Sci Sports Exerc, 36*(7), 1165-1174.

Tsai, L. C., Sigward, S. M., Pollard, C. D., Fletcher, M. J., & Powers, C. M. (2009). Effects of Fatigue and Recovery on Knee Mechanics during Side-Step Cutting. *Med Sci Sports Exerc*.

Wallace, B. J., Kernozek, T. W., Mikat, R. P., Wright, G. A., Simons, S. Z., & Wallace, K. L. (2008). A Comparison between Back Squat Exercise and Vertical Jump Kinematics: Implications for Determining anterior cruciate ligament Injury Risk. *Journal of Strength & Conditioning Research, 22*(4), 1249-1258.

Yeow, C. H., Lee, P. V., & Goh, J. C. (2009). Effect of landing height on frontal plane kinematics, kinetics and energy dissipation at lower extremity joints. *J Biomech, 42*(12), 1967-1973.

Yu, B., Lin, C. F., & Garrett, W. E. (2006). Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech (Bristol, Avon), 21*(3), 297-305.

Zantop, T., Petersen, W., Sekiya, J. K., Musahl, V., & Fu, F. H. (2006). Anterior cruciate ligament anatomy and function relating to anatomical reconstruction. *Knee Surg Sports Traumatol Arthrosc, 14*(10), 982-992.