The relationship between lower extremity muscle activity and knee flexion angle during a jump-landing task.

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ABSTRACT

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jump-landing task.
(Under the direction of Darin A. Padua)

Objective: To determine relationships between lower extremity muscle activity and knee flexion angle during a jump-landing task. **Subjects:** Thirty recreationally active people free from previous ACL injury. **Methods:** Muscle activity of the lower extremity and knee flexion angle were recorded during 10 trials of the jump-landing task. Data was analyzed using correlation and multiple regression statistical analyses. **Results:** Eight significant negative relationships were found, VMO, GMAX activity and Q:H ratio with knee flexion angle at initial contact (VMO: $r = -0.382$, $P = 0.045$, GMAX: $r = -0.385$, $P = 0.043$, Q:H ratio: $r = -.442$, $P = .018$). VMO, VL, and GMAX activity with peak knee flexion angle (VMO: $r = -0.687$, $P = 0.00$, VL: $r = -0.467$, $P = .011$, GMAX: $r = -0.386$, $P = 0.043$). VMO and VL activity with knee flexion displacement (VMO: $r = -0.631$, $P = 0.00$, VL: $r = -0.453$, $P = 0.014$). Q:H ratio and GMAX activity predicted 34.7% of knee flexion angle variance at initial contact ($R^2 = 0.347$, $P = 0.006$). VMO activity predicted 47.1% of peak knee flexion angle variance ($R^2 = 0.471$, $P = 0.000$). VMO and VL activity predicted 49.5% of knee flexion displacement variance ($R^2 = 0.495$, $P = 0.000$). **Conclusions:** Increased quadriceps and GMAX activation, with a lack of hamstring and gastrocnemius activation, showed a relationship with decreased knee flexion angle and we speculate this movement pattern to be a potential risk-factor for ACL injury.

DEDICATION

This project has gone through a lot of the same growth that I have undergone over the past two years. It was not without help that it was completed, and many, many, thanks must go to Michelle Boling, without whom this project would not have been completed, Darin Padua, for all of the guidance and answers to the hard questions, Melanie McGrath, the Vicon expert and also Troy Blackburn for his help with the project. A huge thank you goes to my data collection partner, Lis Macrum, who was vital in collecting enough data to even complete the project. Thank you to all those involved in Carolina Sports Medicine for their help and support in getting this project created, started, and completed. Thank you also to Terri Jo Rucinski and Doug Halverson, allowing me the flexibility within my clinical responsibilities to complete this project.

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CHAPTER 1

INTRODUCTION

A milestone for female athletics was the passage of Title XI in 1972. Since then, participation by females in athletics has grown immensely. Along with the increase in participation has come an increase in injuries in female athletes, notably anterior cruciate ligament (ACL) injuries (Arendt, Agel et al. 1999; Agel, Arendt et al. 2005). It has been shown that females participating in sports requiring pivoting and jumping suffer ACL injuries at a 4-6 times greater rate than males participating in the same sport (Hewett, Myer et al. 2005). An estimated 38,000 ACL injuries occur in females in the United States annually, costing nearly $646 million dollars (Hewett, Myer et al. 2005). Additionally, ACL injuries have lead to increased development of osteoarthritis, knee instability, and decreased academic performance (Freedman, Glasgow et al. 1998; Yu and Garrett 2007). Due to the associated cost and long term disability, it is important to understand potential factors that may contribute to the higher incidence of ACL injury in females compared to males.

Although ACL injury has become a focus of sports medicine research, the explanation behind the prevalence of females suffering non-contact ACL injury has remained largely unclear. Several risk factors for female non-contact ACL injuries have been suggested, focusing on three main etiologic areas: anatomical, hormonal, and neuromuscular (Hewett, Myer et al. 2005). Although anatomical and hormonal differences have been

identified between females and males, these factors remain mainly non-modifiable. An increasing amount of evidence suggests that altered muscle activation and joint positioning during high risk deceleration activities such as running, cutting, pivoting, or landing may be the main contributing factors to female non-contact ACL injury (Malinzak, Colby et al. 2001; Hewett, Myer et al. 2005). Muscle activation patterns and joint positions that directly influence ACL loading are believed to play an important role in the elevated risk of ACL injury observed in females.

The ACL provides approximately 80% of the passive restraint against proximal anterior tibial shear forces (PATSF) and anterior tibial translation (ATT). While other loading conditions (e.g. knee valgus moment, tibial internal rotation moment) have been shown to also cause ACL loading (Fukuda, Woo et al. 2003), PATSF has been shown to be the primary ACL loading mechanism (Markolf, Gorek et al. 1990; Markolf, Burchfield et al. 1995). Minimizing ACL loading induced by PATSF requires appropriate muscle activation of the lower extremity musculature while maintaining a more flexed knee position (Li, Defrate et al. 2005; Shelburne, Torry et al. 2005; Shelburne, Torry et al. 2005). Specifically, the quadriceps and gastrocnemius act as ACL antagonists and may increase PATSF along with resultant ACL loading when the muscles contract (Fleming, Renstrom et al. 2001; Pflum, Shelburne et al. 2004; Padua, Carcia et al. 2005; Shelburne, Torry et al. 2005) Conversely, the hamstrings are an ACL agonist and reduce ACL loading by directly offsetting PATSF (Liu and Maitland 2000; Shelburne, Torry et al. 2005). The ability of these muscles to affect ACL loading is modulated by knee flexion angle (Li, Rudy et al. 1999; Li, Zayontz et al. 2004; Li, Defrate et al. 2005). At small knee flexion angles the ability of the quadriceps and gastrocnemius to induce PATSF and ACL loading is maximized while the

hamstring's ability to offset PATSF loading is diminished (Withrow, Huston et al. 2006; Sell, Ferris et al. 2007). Proximal to the knee, sagittal hip joint motion has been shown to have an affect on PATSF, with lesser hip flexion motion being significantly correlated with greater peak posterior and vertical ground reaction forces (Yu, Lin et al. 2006). Increased peak posterior and vertical ground reaction forces have been shown to be significantly correlated to increased peak PATSF (Yu, Lin et al. 2006).

Several studies have shown females to have different lower extremity motion patterns than their male counterparts during non-contact athletic tasks (Malinzak, Colby et al. 2001; Hewett, Myer et al. 2005; Chappell, Creighton et al. 2007). Specifically, females have been shown to perform athletic tasks with decreased knee flexion angles, increased knee valgus angles, and decreased hip flexion angles (Malinzak, Colby et al. 2001; Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). Females have also been reported to display different muscle activity patterns when compared to males. Particularly, females have shown a trend toward higher quadriceps, soleus, and gastrocnemius activity and lower hamstring activity compared to males (Malinzak, Colby et al. 2001; White, Lee et al. 2003; Padua, Carcia et al. 2005; Kvist 2006). It is hypothesized that the observed sex differences in muscle activation and knee flexion angle during cutting, jumping, landing, and running may explain the sex differences in non-contact ACL injury rates (Hewett, Ford et al. 2006; Yu, Lin et al. 2006; Chappell, Creighton et al. 2007; Sell, Ferris et al. 2007). As previously stated, decreased knee flexion during athletic tasks is believed to be particularly problematic due to the influence of knee flexion angle on the resultant PATSF induced by quadriceps, hamstrings, and gastrocnemius muscle contraction. Thus, when considering the influence of muscle activation on PATSF it is important to also consider knee flexion angle in the sagittal plane.

Muscular co-contraction creates increased joint compression, which allows the load applied to the knee to be distributed evenly over the articular surfaces between the tibia and femur (Hewett, Myer et al. 2005). When the articular surfaces of the knee joint can absorb the forces applied, the forces are not transmitted to the joint's ligamentous restraints and are therefore protected (Hewett, Zazulak et al. 2005). As discussed earlier, the quadriceps, hamstrings and gastrocnemius all influence the stability of the knee (Liu and Maitland 2000; Fleming, Renstrom et al. 2001; Pflum, Shelburne et al. 2004; Shelburne, Torry et al. 2005). Ideally, a balance between ACL agonist muscles (hamstrings) with the antagonist muscles (quadriceps and gastrocnemius) along with proximal knee stabilizers is required to perform athletic tasks while maintaining knee joint stability and minimizing ACL load. However, research has not examined the influence of muscle co-activation patterns on knee flexion angles.

In a recent review by Yu et al (Yu and Garrett 2007), several mechanisms were identified as potential factors in ACL injury. Mainly the authors identified sagittal plane biomechanical factors including small knee flexion angle, large quadriceps force, and large posterior ground reaction force as major ACL loading mechanisms (Yu and Garrett 2007). Sell et al (Sell, Ferris et al. 2007) recently completed a study using a variety of factors to predict proximal ATSF. The authors reported that knee flexion angle and EMG activity of the vastus lateralis and sex (female) significantly predicted PATSF (Sell, Ferris et al. 2007). While these authors were able to suggest important sagittal plane factors that may lead to ACL injury, they failed to explore proximal and distal factors, like muscular activation patterns of the gluteus maximus and gastrocnemius, that may also influence the lower extremity and the resultant loads placed on the ACL. Additionally, the authors fail to explore

the relationship between EMG patterns and knee flexion angle. It has been shown in the literature that both EMG patterns and knee flexion angle influence ACL loading (Sell, Ferris et al. 2007; Yu and Garrett 2007), however the literature lacks a definitive relationship between knee flexion angle and lower extremity EMG patterns. Once defined, this relationship may lead to a greater understanding of the factors that induce ACL loading.

The alarming rate of ACL injury in females has highlighted the importance of prevention as a therapeutic tool. It has been well established in the literature that females tend to have altered lower extremity motion patterns and muscle activity in the quadriceps and hamstrings during athletic tasks (Malinzak, Colby et al. 2001; Hewett, Zazulak et al. 2005; Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). It has also been documented that these neuromuscular differences within the lower extremity may increase PATSF, which predisposes the ACL to injury (Fukuda, Woo et al. 2003; Pflum, Shelburne et al. 2004; Shelburne, Torry et al. 2005). While correction of these biomechanical and neuromuscular faults directly at the knee is a starting point, muscular activation proximal and distal to the knee should not be disregarded. Most athletic tasks involve the coordination of the muscles throughout the entire lower extremity to create movement. However, the literature lacks data describing the collective effects of sagittal plane musculature of the lower extremity on the knee during athletic tasks. Therefore, the purpose of this study is to determine relationships between lower extremity muscle activity (quadriceps, hamstrings, gluteus maximus, and gastrocnemius) and knee flexion angle during a jump-landing task. By determining these relationships, conclusions can be drawn about the effect the muscles have on knee flexion angle and the resultant loads that may be placed on the ACL due to anterior tibial shear force.

These relationships can then be utilized in the development of injury prevention and rehabilitation programs.

Research Questions

1. Is there a relationship between quadriceps EMG amplitude and knee flexion angle during a jump landing task?

2. Is there a relationship between hamstring EMG amplitude and knee flexion angle during a jump landing task?

3. Is there a relationship between gluteus maximus EMG amplitude and knee flexion angle during a jump landing task?

4. Is there a relationship between gastrocnemius EMG amplitude and knee flexion angle during a jump landing task?

5. Does a combination of quadriceps, hamstring, gluteus maximus and gastrocnemius EMG predict knee flexion angle during a jump landing task?

Null Hypothesis

1. H_0 = There is no relationship between quadriceps EMG amplitude and knee flexion angle during a jump landing task.

2. H_0 = There is no relationship between hamstring EMG amplitude and knee flexion angle during a jump landing task.

3. H_0 = There is no relationship between gluteus maximus EMG amplitude and knee flexion angle during a jump landing task.

4. H_0 = There is no relationship between gastrocnemius EMG amplitude and knee flexion angle during a jump landing task.

5. H_0 = The combination of quadriceps, hamstring, gluteus maximus, and gastrocnemius EMG amplitude does not predict knee flexion angle during a jump landing task.

Research Hypothesis

1. There will be a negative relationship between quadriceps EMG amplitude and knee flexion angle during a jump landing task.

2. There will be a positive relationship between hamstring EMG amplitude and knee flexion angle during a jump landing task.

3. There will be a negative relationship between gluteus maximus EMG amplitude and knee flexion angle during a jump landing task.

4. There will be a negative relationship between gastrocnemius EMG amplitude and knee flexion angle during a jump landing task.

5. The combination of quadriceps, hamstring, gluteus maximus and gastrocnemius EMG activity will predict knee flexion angle during a jump landing task.

Operational Definitions

Surface electromyography: Measurement of the electrical activity of muscles using surface electrodes.

Anterior tibial shear force: The amount of force directed anteriorly at the tibiofemoral joint.

Jump landing task: Jumping off a 30 cm high platform with both legs from a horizontal distance equal to 50% of the subject's height from the front edge of a force plate. After landing subjects will be instructed to recoil and perform a rebound jump for maximal vertical height.

Dominant leg: The leg used to kick a soccer ball for maximum distance.

Initial Contact: The first time point during jump landing task the vertical ground reaction force goes exceeds10N.

Deceleration phase: The phase during the jump landing task from initial contact to the time point of peak knee flexion.

Delimitations

1. All subjects will be recreationally active, participating in physical activity 3 days per week for at least 30 minutes.

2. All analyses will be performed on the dominant leg.

3. Subjects will perform the jump-landing task as they would in a sport setting.

Assumptions

1. All subjects will be truthful about the lack of previous lower extremity injury.

2. The jump landing task is a common task that increases the load on the tibia, thus increasing ATSF.

3. Increased ATSF is an accurate predictor of ACL injury.

4. The EMG and kinetic data collected through Motion Monitor software are reliable and valid.

CHAPTER 2

LITERATURE REVIEW

Introduction

Injury to the anterior cruciate ligament (ACL) is a common and devastating injury within the athletic population. Athletes suffering from an ACL injury are often disqualified from participation for several months at a time. ACL injury can occur during contact situations, like collisions with another player, or non-contact situations such as cutting, planting, or landing from a jump. Contact injuries are an inherent risk of sport participation, whereas non-contact injuries can become extremely frustrating. A current trend seen in ACL injury is an increased occurrence in female athletes, especially female non-contact ACL injury (Hewett, Myer et al. 2005). In recent years ACL injury has become a focus of Sports Medicine research. It has been proposed in the literature that females suffer more non-contact ACL injury than males based on anatomical, hormonal, and neuromuscular differences between the sexes. While anatomical and hormonal differences between males and females remain largely non-modifiable, it has been proposed that differences in neuromuscular control can be changed with training programs (Hewett, Ford et al. 2006) . One potential neuromuscular risk for ACL injury is the position of the knee while landing from a jump. It has been suggested that landing

from a jump increases proximal anterior tibial shear force (PATSF) exerted at the knee, due to contact ground reaction forces and the position of the knee while landing (Sell, Ferris et al. 2007; Yu and Garrett 2007). PATSF has been shown to load the ACL, at times with large enough forces to exceed the failure strength of the ligament (Markolf, Gorek et al. 1990; Markolf, Burchfield et al. 1995). Improvements in neuromuscular control through exercises focusing on stabilizing while landing from a jump may help to prevent ACL injuries. Landing from a jump involves coordination of all of the musculature of the lower extremity. The gluteus maximus, quadriceps, hamstrings and gastrocnemius are four of the muscles that influence the forces exerted at the tibiofemoral joint. ACL loading as a result of the contraction of these muscles has been shown to be influenced by knee flexion angle (Li, Rudy et al. 1999; Li, Zayontz et al. 2004; Li, Defrate et al. 2005). Thus the dynamic position of the knee as a result of contraction of lower extremity musculature may lead to heavy load transmission to the ACL. Gaining knowledge about the musculature acting on the tibiofemoral joint, along with the resultant knee flexion angle, may help to develop injury prevention plans in the future. The purpose of this review is to rationalize the investigation of the relationship between the musculature acting on the knee and knee flexion angle when landing from a jump.

Etiology

One of the common injuries sustained by athletes is anterior cruciate ligament (ACL) rupture. ACL rupture is often a devastating injury that causes the athlete to be disqualified for several months at a time. It has been estimated that the annual incidence of ACL injury is one for every 1,750 persons aged 15 to 45 (Griffin, Agel et al. 2000). Both contact and non-contact mechanisms have been found to be the cause of ACL injury

(Agel, Arendt et al. 2005). While contact injuries remain an inherent risk of sport, non-contact injuries have become a frustrating aspect of athletic participation. Several high risk maneuvers for ACL injury, including rapid deceleration, cutting, pivoting, awkward landings and "out of control" play, have been identified in the literature (Griffin, Agel et al. 2000). Sports that involve high levels of these maneuvers, such as soccer, basketball, volleyball and lacrosse, consistently expose athletes to these dangerous positions throughout practice and games.

Although ACL injuries occur both in males and females, it has been shown in the literature that females tend to suffer more ACL injuries when compared to their male counterparts when participating in the same sport (Agel, Arendt et al. 2005; Mihata, Beutler et al. 2006). In a landmark study done in 1995 using the NCAA Injury Surveillance System, Arendt and Dick (Arendt and Dick 1995) investigated knee injuries among collegiate athletes. The authors found females have a significantly higher incidence of ACL injury, regardless of mechanism, than males participating in soccer and basketball. In continuation with Arendt and Dick, Mihata et al (Mihata, Beutler et al. 2006) investigated the rate of ACL injury in female athletes using the NCAA ISS information over a 15 year period from 1989 to 2004, excluding 1996-1997. The authors found females to have three times more soccer ACL injuries and four times more basketball ACL injuries than males, regardless of mechanism. Furthermore, it has also been suggested in the literature that females tend to suffer more non-contact ACL injury than males (Agel, Arendt et al. 2005). Upon review of the NCAA ISS information, Agel et al (Agel, Arendt et al. 2005) reported that females had a significantly higher incidence of non-contact ACL injury than males, regardless of sport. Among the sample of schools

participating in the NCAA ISS collection, non-contact female ACL injury constituted 67% of total ACL injury (Agel, Arendt et al. 2005). With female ACL surgery and rehabilitation estimated to cost the United States $646 annually (Hewett, Myer et al. 2005), it is clear that strategies for prevention should be the goal of researchers today.

Anatomy

In order to complete a research study on the ACL, the complex anatomy of the knee must be understood. The knee is a primarily hinge synovial joint made of the femur, tibia and patella. The knee consists of three articulations, the medial femoral condyle with the tibial femoral condyle, the lateral femoral condyle with the lateral tibial condyle, and the patella with the femur. The articular surfaces of the knee are characterized by their large surfaces and incongruent shapes. Mechanically, the knee is considered weak because of the complicated configurations of the articular surfaces. Therefore, the knee is primarily stabilized by the passive ligamentous structures and the active musculature and associated tendons.

The knee is surrounded by a thin articular capsule. The strong fibrous capsule attaches to the femur superiorly, proximal to the articular margins of the condyles, to the intercondylar fossa posteriorly and to the articular margin of the tibia inferiorly. The fibrous capsule does not line the lateral aspect of the joint to allow the popliteus tendon to pass out of the joint and attach to the tibia. Anteriorly the patella and patellar ligament serve as the capsule. Lining the internal aspect of the fibrous capsule is the synovial membrane, which attaches to the edge of the patella and the edges of the menisci, fibrocartilaginous discs between the femoral and tibial condyles. The synovial membrane runs from the posterior aspect of the joint to the cruciate ligaments and covers the

infrapatellar fat pad and cruciate ligaments anteriorly, separating the cruciate ligaments from the joint cavity.

The knee joint is supported by passive ligamentous structures medially, laterally, anteriorly, and posteriorly. The medial collateral ligament (MCL) is a strong flat band extending from the medial epicondyle of the femur to the superior medial surface of the tibia with deep fibers attaching directly to the medial meniscus. The MCL's primary role is to resist valgus motion about the knee. The lateral collateral ligament (LCL) is a round, cordlike band extending from the lateral epicondyle of the femur to the head of the fibula, and primarily resists varus motion about the knee. The cruciate ligaments, both anterior and posterior, join the tibia and the femur with the articular capsule of the joint but outside the synovial joint cavity. The posterior cruciate ligament (PCL) is the stronger of the two, and runs from the posterior intercondylar area of the tibia to the anterior lateral surface of the medial condyle of the femur. The PCL resists anterior displacement of the femur on the tibia or posterior displacement of the tibia on the femur. The anterior cruciate ligament (ACL) arises from the anterior intercondylar area of the tibia and runs superiorly, posteriorly, and laterally to the medial side of the lateral condyle of the femur. The ACL is the main restraint of anterior tibial movement on the femur or posterior displacement of the femur on the tibia. The ACL is made of two discrete segments, the anteromedial bundle and posterolateral bundle. When the knee is fully extended, the anteromedial bundle is anterior to the posterolateral bundle. As the knee moves into flexion, the positions are reversed, causing the ACL to wind upon itself. This causes varying portions of the ACL to be taut as the knee moves through its normal range of motion.

The knee is also supported actively by the muscles and associated tendons surrounding the joint. The quadriceps form the bulk of the anterior thigh muscles and consists of four parts: rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius. The quadriceps share a common insertion on the tibial tuberosity and are mainly responsible for extending the knee. The hamstrings are located on the posterior aspect of the thigh and consist of the semitendinosus, semimembranosus, and biceps femoris. Collectively the hamstrings serve to extend the hip and flex the knee. Proximal to the knee joint, the gluteus maximus, the most superficial gluteal muscle, extends inferolaterally from the pelvis to the buttock. The main action of the gluteus maximus is extending the thigh, while it also assists in lateral thigh rotation and overall stability of the lower extremity. The gastrocnemius forms the superficial posterior compartment of the lower leg, running from the posterior aspect of the medial and lateral femoral condyles to the calcaneus via the Achilles tendon. The main action of the gastrocnemius is plantarflexion of the ankle, however it assists in knee flexion as well.

Muscle Contraction and ACL Loading

It has been shown in the literature that the resultant forces of the patellar tendon, quadriceps, hamstrings, tibiofemoral contact, and ground reaction acting on the tibiofemoral joint during landing and walking produce force in the anterior direction (Pflum, Shelburne et al. 2004; Shelburne, Torry et al. 2005). This increase in force in the anterior direction will cause an increase in anterior tibial translation, putting increased stress on the ACL. Shelburne et al (Shelburne, Torry et al. 2005), found that a model ACL was loaded whenever the resultant force acting on the knee was in the anterior direction during walking. Pflum et al (Pflum, Shelburne et al. 2004) also found that peak

proximal anterior shear force occurred at the precise time the maximum force was transmitted to the ACL during a simulated landing task. When these high proximal anterior shear forces are produced and transmitted to the ACL during athletic movement, they may exceed the ligament's failure strength and cause rupture.

Quadriceps. Studies done with both cadavers and 3-D models have shown structures acting at the tibiofemoral joint exert shear forces, which may transmit forces to the ACL. The patellar tendon has been shown to apply a proximal anterior shear force to the lower leg at certain knee flexion angles (Pflum, Shelburne et al. 2004; Shelburne, Torry et al. 2005). Using a 3-D model to simulate landing, Pflum et al (Pflum, Shelburne et al. 2004) found the patellar tendon applied a proximal anterior shear force to the lower leg throughout the entire landing task. The proximal anterior shear force produced by the patellar tendon peaked immediately after initial impact (600N ~0.9 body weight). Additional authors found that proximal anterior shear force produced by the patellar tendon dominated all shear forces acting on a lower leg model during early stance of walking (Shelburne, Torry et al. 2005). Agreement among the authors suggested that the large proximal anterior patellar tendon force was a result of the knee being in a more extended position, thus causing a large angle between the patellar tendon and long axis of the tibia (Pflum, Shelburne et al. 2004; Shelburne, Torry et al. 2005).

Considering their attachment with the patellar tendon, it can be concluded the quadriceps also produce a proximal anterior shear force at knee. During the above described 3-D simulated landing, it was found the quadriceps, specifically the vasti, were the prime movers of the lower limb (Pflum, Shelburne et al. 2004). Shelburne et al (Shelburne, Torry et al. 2005) were also able to demonstrate that peak patellofemoral

force coincided with the appearance of peak quadriceps force during normal walking. In a recent study by Demorat et al (DeMorat, Weinhold et al. 2004), the authors were able to demonstrate that a 4500-N quadriceps force could produce ACL injuries in cadaveric knees at 20° of knee flexion. Eleven cadaver knees were fixed and loaded with a 4500-N quadriceps force; six of the eleven knees had confirmed ACL injury and all eleven demonstrated increased anterior tibial translation (ATT). The results of these studies confirm the large influence the quadricep muscles have on ATT and potentially increasing ACL loading.

Hamstrings. In contrast to the patellar tendon and quadriceps, the hamstrings have been shown to apply posterior shear force to the lower leg. Pflum et al (Pflum, Shelburne et al. 2004) found that the hamstrings applied a posterior force to the lower leg in a 3-D model throughout an entire simulated landing task. The authors also found the force produced by the hamstrings increased with time, reaching peak around the end of the landing phase. Also using a 3-D model, Shelburne et al (Shelburne, Torry et al. 2005) found that normal levels of anterior tibial translation (ATT) could be restored in ACL deficient knees by increasing the magnitude of force produced in the posterior direction by the hamstrings. Li et at (Li, Rudy et al. 1999) demonstrated that an antagonistic hamstring contraction reduced ATT at knee flexion angles between 30° and 120°. In addition, the antagonistic hamstring contraction reduced in-situ forces to the ACL at 15°, 30° and 60° of flexion (Li, Rudy et al. 1999). Increased activity within the hamstrings has long been thought to resist ATT and potentially result in reduced forces being transmitted to the ACL.

Gastrocnemius. Several studies using models have shown the gastrocnemius to apply a small proximal anterior tibial shear force because of their anatomical contact with the posterior femoral condyles (Pflum, Shelburne et al. 2004; Shelburne, Torry et al. 2005). While the exact mechanism of gastrocnemius ACL loading is unknown, it is hypothesized that the anatomical attachment of the gastrocnemius creates a posterior translation of the femur relative to the tibia (Padua, Carcia et al. 2005). Recent research has demonstrated the gastrocnemius as an antagonist to the ACL by increasing ACL strain (Fleming, Renstrom et al. 2001). Fleming et al, (Fleming, Renstrom et al. 2001), demonstrated that isolated gastrocnemius contraction increased ACL strain values, especially with the knee near full extension. Further, the authors demonstrated that co-contraction of the gastrocnemius and quadriceps produced even higher ACL strain values (Fleming, Renstrom et al. 2001). However, arguments have been made that co-contraction of the quadriceps and gastrocnemius may cause joint compression that may aid in decreasing anterior tibial shear force during closed chain activities (Kvist 2006). The literature lacks a clear definition of the effect of the gastrocnemius on ACL loading.

Gluteus maximus. While the gluteus maximus has no direct action on the knee, activity within the hip musculature affects lower extremity alignment and the load bearing capacity of the knee (Hewett, Zazulak et al. 2005). Recently Yu et al (Yu, Lin et al. 2006) were able to show that hip joint angular velocity affected proximal anterior tibial shear force during a stop-jump task. The results of their study showed that hip angular velocity was significantly correlated with peak posterior ground reaction force, and peak posterior ground reaction force was significantly correlated with peak proximal anterior tibial shear force (Yu, Lin et al. 2006). Specifically the authors were able to

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show that decreased hip motion led to increased PATSF (Yu, Lin et al. 2006).

Conclusions can be made from this study that hip movement affects the forces applied to

the knee through PATSF, which may put increased stress on the ACL.

Influence of Knee Flexion Angle on ACL Loading

The musculature at and surrounding the knee have been shown to affect loads

transmitted to the ACL through PATSF. The quadriceps have been shown to be the

major contributor to PATSF via the attachment at the patellar tendon (Yu and Garrett

2007). However, the ability of the quadriceps to produce a large PATSF is largely

regulated by the knee flexion angle. The force produced by the quadriceps that loads the

ACL is dependent on the patella tendon-tibia shaft angle, the angle between the patella

tendon and the longitudinal axis of the tibia. With a set quadriceps force, the larger the

patella tendon-tibia shaft angle, the larger the anterior shear force on the tibia (Yu and

Garrett 2007). The relationship between patella tendon-tibia shaft angle and knee flexion

angle has been previously investigated. Nunley et al (Nunley 2003) were able to show

that patella tendon-tibia shaft angle was a function of knee flexion angle and that as knee

flexion angle decreased, patella tendon-tibia shaft angle increased.

Knee flexion angle has also been shown to affect ACL elevation angle, the angle

between the longitudinal axis of the ACL and tibial plateau, as well as ACL deviation

angle, the angle between the projection of the longitudinal axis of the ACL on the tibial

plateau and the posterior direction of the tibia (Li, Defrate et al. 2005). The greater the

ACL elevation and deviation angles the greater the ACL loading is with a set proximal

anterior shear force (Yu and Garrett 2007).

ACL loading has been shown to increase as the knee flexion decreases. Li et al (Li, Rudy et al. 1999; Li, Zayontz et al. 2004) studied the effects of quadriceps and hamstring loading on ACL loading. The authors were able to show that ACL loading increased as knee flexion angle decreased with a quadriceps muscle load, regardless of the load on the hamstrings (Li, Rudy et al. 1999; Li, Zayontz et al. 2004). Considering the current literature it can be concluded that knee flexion angle plays a large roll in regulating potentially large PATSF that may place large loads on the ACL.

ACL Risk Factors

Recently, sports medicine research has been heavily focused on female ACL injury. The epidemiologic information has shown female athletes to be a high risk population for ACL injury (Agel, Arendt et al. 2005; Mihata, Beutler et al. 2006), yet the literature lacks a clear and definitive consensus on female ACL injury. It has been proposed in the literature that female ACL injury can be related to three major risk factors: anatomical, hormonal, and neuromuscular (Hewett, Myer et al. 2005). While clinicians have little control over modification of anatomical and hormonal factors, neuromuscular control may be modified through rehabilitation or prevention programs. Each risk factor for female ACL injury will be discussed further below.

Researchers have linked several differences in female anatomy to being potential risk factors for ACL injury. For example, females have been shown to have a larger Q angle, the acute angle between the line connecting the anterior superior iliac spine to the middle of the patella and the line connecting the tibial tuberosity to the center of the patella (Hughes and Watkins 2006). A larger Q angle inherently puts the knee in a larger valgus angle during movement, which has been associated an increased risk for ACL

injury (Hewett, Myer et al. 2005). It has also been theorized that the size of the femoral notch, the area between the two femoral condyles, is a risk factor for ACL injury. The smaller the size of the femoral notch, which occurs with decreased size of the femur, the less area for the ligaments. In a prospective study done by Uhorchak (Uhorchak, Scoville et al. 2003), 711 male and 113 female cadets were measured at the start of a 4-year period for intercondylar notch width. During the 4 years, 24 non-contact ACL injuries occurred, 16 to males and 8 to females. The results of the study showed females suffering ACL injury had significantly smaller femoral notch width (mean 15.6mm) when compared to males (mean 17.7mm) (Uhorchak, Scoville et al. 2003). It has also been proposed that increased femoral anteversion, excessive tibial torsion, excessive foot pronation, and decreased ACL size in females may lead to increased risk for ACL injury in females, but these factors have yet to be fully explored (Griffin, Agel et al. 2000).

Hormonal risk factors for female ACL injury have also been suggested in the literature. Specifically, it has been suggested that the female menstrual cycle may affect the mechanical properties of the ACL. Differences among the genders in the mechanical properties of the ACL have been demonstrated in cadavers (Chandrashekar, Mansouri et al. 2006). In a study of 10 male and 10 female cadaveric knees taken to ligamentous failure, the authors were able to demonstrate the female ACL had lower mechanical properties when compared to males. In particular, the female ACL had lower stiffness and modulus of elasticity, lower stress at failure, and lower strain density at failure (Chandrashekar, Mansouri et al. 2006). The results of this study support the theory that there may be gender based differences in the modeling of the ACL, which may influenced by the female menstrual cycle. In a meta-analysis performed by Zazulak et al

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(Zazulak, Paterno et al. 2006), nine studies were reviewed and menstrual phase was found to have a significant effect on anterior knee laxity. Anterior knee laxity was found to be the highest during the ovulatory phase of the menstrual cycle, when estrogen levels were the highest, and the lowest during the follicular phase when estrogen levels were the lowest. This meta-analysis supports the theory that female sex hormones influence ligament laxity, which may predispose females to ACL injury. However, a significant link between female ACL injury and menstrual cycle phase has yet to be established in the literature.

Increasing evidence in the literature suggests that poor or abnormal neuromuscular control of the lower leg during dynamic movements is the main contributor to female ACL injury (Malinzak, Colby et al. 2001; White, Lee et al. 2003; Hewett, Myer et al. 2005; Zazulak, Ponce et al. 2005). Neuromuscular control refers to the unconscious activation of dynamic restraints surrounding joints in response to sensory stimuli (Griffin, Agel et al. 2000). While anatomical and hormonal factors remain largely non-modifiable, neuromuscular control has been shown to improve after training (Hewett, Ford et al. 2006). Hewett et al (Hewett, Ford et al. 2006) found that neuromuscular training likely alters active knee joint stability and appears to aid in decreasing ACL injury rates in females. If high risk athletes can be identified through a pre-season screening and then begin a neuromuscular training program, efforts at preventing season-ending ACL injury can start. Neuromuscular control involves several components, including motion patterns and muscle activity, each of which will be explored further.

Lower extremity motion patterns

Important aspects of neuromuscular control are motion patterns and joint angles throughout movement. Several authors have shown females to have different movement patterns during athletic tasks when compared to their male counterparts (Malinzak, Colby et al. 2001; Chappell, Creighton et al. 2007). It has been shown in the literature that females tend to have decreased knee flexion angles during selected athletic tasks. Malinzak et al (Malinzak, Colby et al. 2001) completed a study comparing lower extremity movement patterns of 11 male and 9 female recreational athletes during running, cross-cutting, and a side-cut. The authors were able to show significant differences in knee flexion-extension motion between female and males. Females were shown, in general, to have 8º less knee flexion throughout all three motions, with the maximum difference of 15º seen at takeoff of the foot from the ground during running and side-cutting (Malinzak, Colby et al. 2001). Chappell et al (Chappell, Creighton et al. 2007) investigated 17 male and 19 female healthy recreational athletes during a vertical stop jump task. Females were seen to have 17º of knee flexion at landing whereas males landed in 24º of knee flexion (Chappell, Creighton et al. 2007).

This decrease in knee flexion angle throughout athletic tasks frequently puts the athlete in a high risk position for ACL injury. As described earlier, the smaller the knee flexion angle, the larger the angle between the patellar tendon and the long axis of the tibia. This large patellar tendon angle enables the quadriceps to create a large proximal anterior shear force that must be resisted by the ACL. Additionally, Li et al (Li, Defrate et al. 2005) completed an in vivo study and showed the ACL to be the longest during full extension and to decrease in length as knee flexion angles increase. This study supports

the theory that the ACL plays its most important role in resisting anterior tibial translation (ATT) during low knee flexion angles. Due to the face that the ACL is relied on most during low knee flexion angles, this is the position where the ligament is most susceptible to injury.

In addition to decreased knee flexion angles, knee valgus angles have been associated with ACL injury risk. Fukuda et al (Fukuda, Woo et al. 2003) showed that anterior tibial translation increased 75-291% at all levels of valgus torques in ACL deficient cadaveric knees, with the largest increases found near full extension. This study highlighted the role the ACL plays in resisting valgus forces at the knee. Several authors have described females to have increased valgus angles, and thus related valgus forces, during athletic tasks (Malinzak, Colby et al. 2001; Hewett, Myer et al. 2005). Malinzak et al found females were consistently in valgus throughout running, cross-cutting, and side-cutting. The authors found females to have 11° more valgus movement about the knee than males throughout the entire movement cycle (Malinzak, Colby et al. 2001). In a prospective study completed in 2003, increased knee valgus angles during a jump landing task were found to be a significant predictor of ACL injury in a group of high school female athletes (Hewett, Myer et al. 2005). Two hundred and five adolescent female soccer, basketball, and volleyball players were screened using a 3D biomechanical analysis of a jump landing task prior to their sport seasons. During the two winter seasons and one fall season the athletes were followed, nine ACL injuries occurred. The nine ACL injuries were confirmed in the injured subjects and their biomechanical analysis was compared to the remaining athletes, who were considered "uninjured". Athletes suffering ACL injury had significantly larger knee valgus angles at

initial contact with the ground, 8.4°, than uninjured athletes. There was also a trend of decreased knee flexion angles seen in ACL injured athletes, 10.5° less at initial ground contact (Hewett, Myer et al. 2005). This prospective study has helped to demonstrate that altered motion patterns may lead to ACL injury in females.

In addition to altered motion patterns about the knee, differences in hip motion patterns have been seen in females when compared to males. Although some may not consider hip motion in relation to knee injury, it is important to appreciate the hip's role in control of lower extremity alignment (Hewett, Zazulak et al. 2005). Chappell et al (Chappell, Creighton et al. 2007) were able to show altered hip motion patterns in females when compared to males during a vertical stop jump task. The authors found decreased hip flexion angles, 48° in females and 56° in males, at landing. It was also found that females had increased internal rotation and abduction-adduction movement during the flight phase prior to landing. Female subjects had their hips internally rotated 9° during the flight phase whereas males were in 14° of hip external rotation throughout the phase. During the landing, female subjects had their hip abducted at 10° whereas males were in 12° of hip abduction (Chappell, Creighton et al. 2007).

Motion patterns around the hip throughout athletic movement have been thought to affect ACL loading. Yu et al (Yu, Lin et al. 2006) demonstrated that hip kinematics affected lower extremity kinetics during a landing task. The authors were able to show that peak posterior ground reaction force during landing was significantly correlated with hip flexion velocity. The authors also demonstrated that peak posterior ground reaction force significantly affected peak proximal anterior tibial force. The results of this study concluded that active hip motion at initial foot contact with the ground reduced impact

force during landing (Yu, Lin et al. 2006). Within the study, it was also demonstrated that females showed decreased hip motion when compared to males during the landing task (Yu, Lin et al. 2006). These decreases in hip motion may predispose the ACL to increased amounts of force when landing from a jump. Additionally, hip adduction and internal rotation motions have been correlated with knee valgus angles (Hewett, Myer et al. 2005). As discussed earlier, increased knee valgus angles have been identified as a high risk position for ACL injury. The results of these studies demonstrated that motion at the hip will affect the motion and force transmission at the knee. It has been found that females tend to have decreased hip flexion angles, which may increase the load placed on the ACL during an athletic task.

Muscle activity of the lower extremity

A second component of neuromuscular control is muscle activity that may influence motion at the knee joint. Non-contact ACL injury occurs during high dynamic loading situations when the active restraints, the musculature influencing the knee joint, fail to dampen the loads and the passive ligaments are subjected to increased stress (Hewett, Zazulak et al. 2005). Many researchers have theorized that muscular control of the lower extremity may prevent altered knee motions (Hewett, Zazulak et al. 2005). However, it has been previously demonstrated that females have altered muscle activity patterns when compared to males (Malinzak, Colby et al. 2001; Hewett, Zazulak et al. 2005; Zazulak, Ponce et al. 2005; Chappell, Creighton et al. 2007). It has been further theorized that a lack of dynamic neuromuscular control of the lower extremity may be a predisposing factor for ACL injury in females (Hewett, Zazulak et al. 2005).

The quadriceps have been shown in models to be the prime movers of the lower extremity during a landing task, a common deceleration task during athletic movement (Pflum, Shelburne et al. 2004). The anatomical attachment of the quadriceps to the patellar tendon produces proximal anterior tibial shear force in addition to producing tibiofemoral contact forces. Quadriceps activity in females during athletic activities has been previously investigated. Malinzak et al (Malinzak, Colby et al. 2001) demonstrated female recreational athletes had higher quadriceps activation during running, cross-cut, and side-cut activities when compared males. Chappell et al (Chappell, Creighton et al. 2007) showed females prepared for landing with 12% higher quadriceps activation when compared to males. Muscle activation levels are major indicators of muscle contraction force, therefore increased quadriceps activity may lead to increased quadriceps force which would increase ACL loading (Chappell, Creighton et al. 2007).

In addition to differences in quadriceps activity levels, females have been shown to have differences in hamstring activation levels. While the quadriceps produce proximal anterior tibial shear force, the hamstrings have been shown to apply posterior shear force to the lower leg in models (Pflum, Shelburne et al. 2004; Shelburne, Pandy et al. 2004). Malinzak et al (Malinzak, Colby et al. 2001) demonstrated decreased hamstring activity in females when compared to males during three common athletic tasks. Additionally, Chappell et al (Chappell, Creighton et al. 2007) showed decreased hamstring activity in females when compared to males following a landing from a jump. While decreased hamstring activation alone may not necessarily increase the possibility of proximal anterior tibial shear force, when combined with increased quadriceps activity, the likelihood of higher levels of proximal anterior tibial shear force is

augmented (Malinzak, Colby et al. 2001). Female subjects have consistently demonstrated higher levels of quadriceps activation and decreased hamstring activation, potentially predisposing the ACL to increased loads.

While the musculature with direct attachment to the tibiofemoral joint certainly causes action about the knee joint, it is important to appreciate the influence of the proximal stabilizing muscles. Decreased activity of the hip musculature may lead to lower extremity malalignment and decrease the load bearing capacity of the knee joint (Hewett, Zazulak et al. 2005). Zazulak et al (Zazulak, Ponce et al. 2005) reported lower gluteus maximus activity in female Division I athletes when compared to male Division I athletes during the postcontact phase of landing. Yu et al (Yu, Lin et al. 2006) were able to demonstrate that hip motion during landing is related to peak ground reaction forces. The authors were able to demonstrate that increased motion about the hip is associated with decreased impact forces of landing from a jump (Yu, Lin et al. 2006). In this study, females were shown to have smaller amounts of hip motion when compared to males (Yu, Lin et al. 2006). This hip motion is thought to be controlled by the musculature surrounding the hip, including the gluteus maximus. In agreement with these studies, it can be concluded that landing with increased hip flexion, as controlled by gluteal muscles, may assist in reducing ACL loading (Chappell, Creighton et al. 2007).

The literature has shown that contraction of the gastrocnemius increases ACL strain values, but has also argued that this contraction may play a role in stability within the knee joint (Fleming, Renstrom et al. 2001; Kvist 2006). Fleming et al (Fleming, Renstrom et al. 2001) demonstrated that co-contraction of the gastrocnemius and quadriceps increased ACL strain values. Padua et al (Padua, Carcia et al. 2005) found

that women showed a trend towards increased gastrocnemius activity over men (25%

greater than men) during double-leg hopping. In the same study, women were found to

have greater quadriceps activity performing the hopping task over men (Padua, Carcia et

al. 2005). Conclusions can be drawn from this study to show that women have increased

quadriceps and gastrocnemius activity which may place an increased strain on the ACL.

Neuromuscular Training

Altered neuromuscular control of the lower extremity during dynamic movement

has been proposed as a leading factor in female ACL injury (Malinzak, Colby et al. 2001;

Hewett, Zazulak et al. 2005; Hughes and Watkins 2006; Yu, Lin et al. 2006). Many

researchers have proposed that altered neuromuscular control of the lower extremity can

be improved with exercise (Griffin, Agel et al. 2000; Hewett, Ford et al. 2006). Hewett

et al (Hewett, Ford et al. 2006) completed a meta-anaylsis of six articles that were

identified as studies based on training interventions to prevent ACL injury. The authors

were able to show that 3 of 6 neuromuscular training interventions significantly

decreased ACL injuries in females. It was shown in this review that multiple

neuromuscular training components may provide some level of ACL injury risk

reduction. Neuromuscular training likely aids in altering active knee joint stabilization,

which appears to aid in decreasing ACL injury in females (Hewett, Ford et al. 2006).

This review showed that the most effective and efficient programs included a

combination of plyometric training, biomechanical analysis, technique training, balance

and core stability training, and strength training, completed more than once a week for six

weeks.

Jump Landing Task

One of the most frustrating aspects of female ACL injury is the increased non-contact occurrence (Agel, Arendt et al. 2005). Etiologic information has shown 67% of female ACL injury occurred in a non-contact situation (Agel, Arendt et al. 2005). It has been suggested that the most at-risk situation for non-contact ACL injury is deceleration, which may occur as an athlete cuts, changes direction, or lands from a jump (Griffin, Agel et al. 2000). Because of the similarity to a common athletic task, the jump landing task has been demonstrated in several studies when completing research on lower extremity neuromuscular control (Pflum, Shelburne et al. 2004; Hewett, Myer et al. 2005; Hewett, Zazulak et al. 2005; Chappell, Creighton et al. 2007).

The jump landing task is a high risk maneuver for injury due to the increased proximal anterior tibial shear forces created that the ACL must resist. In a simulated 3-D model landing task, the ACL was loaded during the first 25% of the landing phase (Pflum, Shelburne et al. 2004). During the simulated landing, the ground reaction force applied a significant posterior shear force to the lower leg, however total shear applied to the lower leg was directed in the anterior direction (Pflum, Shelburne et al. 2004). The peak proximal anterior shear force occurred precisely at the same instant the maximum force was transmitted to the ACL (Pflum, Shelburne et al. 2004). The resultant proximal anterior shear force was a result of a large eccentric force produced by the quadriceps musculature in order to decelerate the lower limb during the landing (Pflum, Shelburne et al. 2004). While the hamstrings did apply a posterior shear force during the landing, it did not peak until the end of the landing task (Pflum, Shelburne et al. 2004).

Gender differences during jump landing tasks have been previously reviewed (Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). It has been shown that females tend to land with larger impact forces, smaller hip flexion angles, smaller knee flexion angles, increased quadriceps activation, and decreased hamstring activation (Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). The results of these studies clearly demonstrate the gender differences that may lead to increased proximal anterior tibial shear forces during a landing task in females.

Summary

While many researchers have investigated female kinematics, kinetics, and muscle activity during a jump landing task, the literature lacks a definitive relationship between the musculature of the entire lower extremity and knee flexion angle. It has been previously demonstrated that the gluteus maximus, quadriceps, hamstrings, and gastrocnemius muscle all affect knee joint mechanics (Malinzak, Colby et al. 2001; Hewett, Zazulak et al. 2005; Zazulak, Ponce et al. 2005; Kvist 2006; Chappell, Creighton et al. 2007). It has also been demonstrated that females tend to execute athletic tasks with altered lower extremity motion patterns that may lead to increased proximal anterior tibial shear force (Malinzak, Colby et al. 2001; Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). However, the literature lacks a significant model for predicting knee flexion angle based on the activity of the musculature of the lower extremity. Therefore, the purpose of this study is to determine the relationships of lower extremity muscle activity and knee flexion angle during a jump landing task. By determining these relationships, conclusions can be made about the effect the muscles may have on loads placed on the

ACL. These relationships can then be used to model rehabilitation and prevention plans

for ACL injury.

CHAPTER 3

METHODOLOGY

Subjects

Thirty subjects (15 males, 15 females) between the ages of 18-30 years were recruited from The University of North Carolina at Chapel Hill campus. Power analysis for a regression model indicated that for an effect size of 0.8 approximately 30 subjects were necessary. Inclusion criteria included: 1.) participation in physical activity for 30 minutes a day, a minimum of 3 days a week 2.) current (intramural or club) or former (at least one year of high school varsity) participation in organized soccer, volleyball, basketball, or lacrosse. Subjects were free from lower extremity injury in either leg within the past 6 months. Lower extremity injury was defined as any injury sustained resulting in more than one day lost in physical activity or referral to a physician. Subjects were also excluded if they had a history of surgery to the lower extremity within the past 2 years or a history of ACL surgery.

Instrumentation

A surface electromyography (EMG) system (Delsys Bagnoli-8, Boston, MA) was worn by the subject and used to record muscle activity of the gluteus maximus, quadriceps (vastus medialis oblique and vastus lateralis), hamstrings (biceps femoris), and lateral gastrocnemius, via surface electrodes. Raw EMG was collected via Vicon Nexus Software (Vicon Motion Systems, Centennial, CO) and stored for analysis. Unit

specifications for the EMG system included a CMRR of 92 dB and amplifier gain of 1000. A force plate (Bertec Corportation, Columbus, OH) was used to collect kinetic data. Force plate and EMG data were collected at 1200Hz. Each subject was outfitted with reflective markers to record kinematic data of the lower extremity during the jump landing task. The movement of the reflective markers was captured by 7 infrared video cameras (Vicon Motion Systems, Centennial, CO) at a frame rate of 120 Hz. All data were collected using Vicon Nexus Software (Vicon Motion Systems, Centennial, CO). A global reference system was defined using the right hand rule, in which the x-axis is positive in the anterior direction, the y-axis is positive to the left of each subject, and the z-axis is positive in the superior direction

Procedures

Subjects reported to the Sports Medicine Research Laboratory at the University of North Carolina at Chapel Hill for a single testing session lasting approximately 1.5 hours. Subjects were required to wear athletic shoes, and Lycra spandex shorts and shirts were provided to each subject. Upon arrival all subjects read and signed a consent form approved by The University of North Carolina Biomedical Institutional Review Board. Demographic information was collected for each subject and a health questionnaire was used to assess lower extremity injury status. Subjects then completed a 5-minute warm-up on a stationary cycle ergometer at a self-selected pace.

The dominant leg was defined as the leg used to kick a ball for maximum distance and was used for kinematic and EMG data collection for each subject. For EMG preparation, each subject's skin was shaved, abraded and cleaned with isopropyl alcohol prior to application of surface electrodes. The electrodes for the gluteus maximus were

placed over the greatest prominence of the middle of the buttocks and midway between the sacral vertebrae and the greater trochanter (Basmajian, Blumenstein, and Dismatsek, 1980). The electrodes for the hamstrings were placed over the biceps femoris near the measured midpoint of the muscle belly (Basmajian, Blumenstein, and Dismatsek, 1980). The electrodes for the quadriceps were placed over the vastus lateralis, inferolateral to 50% of the quadriceps length (Basmajian, Blumenstein, and Dismatsek, 1980). The electrodes for the gastrocnemius were placed over the bulge of the lateral head of the gastrocnemius (Basmajian, Blumenstein, and Dismatsek, 1980). Electrode placement was confirmed with manual muscle testing. Subjects were then outfitted with reflective markers placed on the following landmarks: Right and left acromion processes, right and left anterior superior iliac spines, S1 joint space, right and left greater trochanters, lateral aspects of the right and left thighs, lateral epicondyle of the right and left knee, medial epicondyle of the right and left knee, lateral aspects of the right and left shanks, right and left lateral malleoli, right and left medial malleoli, right and left heels, the heads of the right and left 5th metatarsals, and the head of the right and left 1st metatarsals. The markers were affixed to the skin and shoes with adhesive tape. Following marker placement, the participant was asked to stand in the center of the calibration area (2.5 m high × 2.5 m long × 1.5 m wide) with each foot on a Bertec Forceplate (Type 4060-08, Bertec Corporation, Worthington, OH), in order to collect a static calibration trial. Following the static calibration trial the medial malleolus and epicondyle markers were removed for data collection during the jump landing task.

EMG activity of the gluteus maximus, quadriceps, hamstrings, and gastrocnemius, and kinematic data were collected during 10 trials of a jump landing task.

The task was first described then demonstrated to the subject. For the task, a 30cm box was placed a distance of 50% of subject's height from the force plate. Subjects were instructed to jump down from the box directly onto the force plates, landing with one foot in the middle of each force plate, then jump straight up for maximum height and land with both feet back on each force plate. The jump-landing task was similar to those previously investigated (Hewett, Myer et al. 2005). Subjects were allowed a maximum of 5 practice trials prior to data collection. A one minute rest period between each of the test trials was given to each subject to avoid fatigue.

Following the data collection trials, each subject performed three maximum voluntary isometric contractions (MVIC) for the gluteus maximus, quadriceps, hamstrings, and gastrocnemius. Each trial was held for 5 seconds. For the gluteus maximus MVIC, the subject was positioned on a table prone, knee flexed to 90 degrees, with a strap placed over the mid-belly of the hamstrings. Subjects were instructed to contract isometrically into hip extension. For MVIC of the quadriceps and hamstrings, subjects were positioned sitting with hips and knees at 90 degrees in a Kincom chair with straps around their legs and trunk. For quadriceps MVIC, subjects were instructed to kick into the strap, extending at the knee. For hamstrings MVIC, subjects were instructed to kick back into the strap, flexing the knee. For gastrocnemius MVIC, subjects laid on the table prone with knees fully extended. A strap was placed around the metatarsal heads of the subject and the subjects were instructed to push into the strap with maximum force. These testing positions are similar to the manual muscle tests described by Hislop and Montgomery (Hislop, Montgomery, 2002).

Data Reduction

Peak knee flexion angle and vertical ground reaction force were used to define the phases of the jump-landing task. Initial contact was defined as the time point in which the vertical ground reaction force exceeded 10N. The preactivation phase was defined as 200ms prior to initial contact. The deceleration phase was defined as the phase from initial contact to the point of peak knee flexion. The stance phase was defined as the phase from initial contact to takeoff. Takeoff was defined as the time point in which vertical ground reaction force was less than 10N.

All EMG data were passively demeaned, bandpass filtered between 10-350Hz, notch filtered between 59.5-60.5Hz, and smoothed using a 20ms root mean squared sliding window function. Mean EMG amplitude of the gluteus maximus, quadriceps, hamstrings, and gastrocnemius were calculated separately for each of the jump-landing trials for the preactivation, deceleration, and stance phases. All EMG data was normalized to the maximum voluntary isometric contraction (MVIC) for each subject during each phase. The mean EMG amplitude was determined during the middle three seconds of each of the five second MVIC trials and the trials were averaged. Normalized EMG was used for data analysis. The quadriceps:hamstring ratio (Q:H ratio) was calculated for each of the phases by calculating the arithmetic mean of the normalized VMO and VL EMG data and then dividing by the normalized hamstrings EMG data.

Knee flexion angle at initial contact was calculated for each subject for each of the ten trials of the jump-landing task. Peak knee flexion angle and knee flexion displacement during the deceleration and stance phases of the jump-landing task were calculated for each subject for each of the ten trials. Knee flexion displacement during

the deceleration and stance phases were determined by subtracting the knee flexion angle at initial contact from the peak knee flexion angle during each phase. The arithmetic mean of knee flexion angle at initial contact was calculated for each subject across the ten trials of the jump-landing task. The arithmetic mean of peak knee flexion angle and displacement were calculated during the deceleration and stance phases across the ten trials of the jump-landing task. Mean knee flexion values were used for statistical analysis.

Statistical Analysis

All data analyses were performed using SPSS version 13.0 (SPSS, Inc. Chicago, IL). Kinematic and EMG data were averaged across the 10 trials for each participant. The reduced kinematic and EMG data collected for each subject were analyzed using correlation and regression analysis to determine relationships between EMG data and knee flexion angle and displacement during each phase of the jump-landing task. Individual simple correlational analyses were run for the relationship between knee flexion angle and displacement and EMG mean amplitude for the gluteus maximus, quadriceps, hamstrings, and gastrocnemius during each phase of the jump-landing task. The variables that were found to have significant correlations were then entered into a forward stepwise multiple linear regression analysis used to predict knee flexion angle and displacement during each phase of the jump-landing task. Variables entered into the regression models were entered in order of highest significance. A priori alpha level was set at 0.05.

CHAPTER 4

RESULTS

Correlation Analysis

Fifteen male (age= 22.2 ± 1.78 years, height = 183.36 ± 6.92 cm, mass = 82.21 ±

11.91 kg) and fifteen female subjects (age= 21.07 ± 2.12 years, height = 164.53 ± 7.38

cm, mass = 62.93 ± 8.91 kg) completed testing for this investigation. Six subjects had

trials that were not included in the statistical analysis due to the EMG data and kinematic

data being outliers (above three standard deviations from the mean). The data that were

not included in the statistical analysis are as follows: subject 1, EMG data of vastus

medialis oblique (VMO) at preactivation and stance phases and all EMG data of the

gluteus maximus; subject 2, all biceps femoris EMG data; subject 9, knee flexion data at

initial contact; and subject 31, EMG data of the VMO during the deceleration phase.

Additionally, subject 4 only had 8 trials of biceps femoris EMG data during the

deceleration phase that were analyzed due to the trial data being above three standard

deviations from the mean; subject 27 had only 9 trials of vastus lateralis EMG data

during both the preactivation and stance phase analyzed due to the trial data being above

three standard deviations from the mean.

Eight significant correlations were found (*P*<0.05) (Tables 4-6). During the

preactivation phase the EMG activity of the VMO and the gluteus maximus (GMAX) and

the Q:H ratio were found to have significant negative relationships with knee flexion angle at initial contact (VMO: r = -0.382, P = 0.045, GMAX: r = -0.385, P = 0.043 ,Q:H ratio: r = -.442, P = .018). During the deceleration phase EMG activity of the VMO, vastus lateralis (VL) and GMAX were found to have significant negative relationships with peak knee flexion angle during the stance phase (VMO: r = -0.687, P =0.00, VL: r = -0.467, P = .011, GMAX: r = -0.386, P = 0.043). Also, during the deceleration phase EMG activity of the VMO and VL demonstrated significant negative relationships with knee flexion displacement during the stance phase (VMO: r = -0.631, P= 0.00, VL: r = -0.453, P = 0.014). No significant correlations were found between biceps femoris or lateral gastrocnemius EMG activity during the preactivation or deceleration phases and knee flexion angle at initial contact, peak knee flexion angle or knee flexion displacement during the stance phase (P >.05) (Tables 4-6). The negative relationships revealed between knee flexion and various measures of EMG activity indicate that decreased knee flexion angle and displacement were associated with increased EMG activity of the VMO, VL, and GMAX muscles and Q:H ratio.

Regression Analysis

The dependent variables that were found to have significant simple correlations were included in the regression analysis (Table 7). A forward stepwise multiple linear regression analysis was used including R^2 change statistics calculated. The regression models presented are those with the highest R^2 change significance (Table 7). Q:H ratio and EMG activity of the GMAX during the preactivation phase were found to predict approximately 34.7% of the variance in knee flexion angle at initial contact (R^2= 0.347, P = 0.006) (Figure 1). During the deceleration phase EMG activity of the VMO was found

to predict approximately 47.1% of the variance in peak knee flexion angle during the stance phase (R^2= 0.471, P = 0.000) (Figure 2). Finally, EMG activity of the VMO and VL during the deceleration phase were shown to predict approximately 49.5% of the variance in knee flexion displacement during the stance phase (R^2= 0.495, P = 0.000) (Figure 3).

CHAPTER 5

DISCUSSION

The purpose of this study was to explore the relationships between muscle activity and knee flexion angle during a jump-landing task in recreationally active males and females. The primary findings of this investigation support a negative relationship between muscle activity of the quadriceps and gluteus maximus musculature and knee flexion angle and displacement during a jump-landing task. In support of our original hypothesis we observed that increased activation of the quadriceps and gluteus maximus musculature was related to a decrease in sagittal plane motion of the knee during a jump-landing task. When assessing the co-activation ratio of quadriceps and hamstrings activity along with gluteus maximus muscle activity during the jump-landing task, we found that preactivation of these muscles predicted knee flexion angle at initial contact. Also, we found that quadriceps muscle activity during the deceleration phase predicted nearly half of the variance in peak knee flexion angle and knee flexion displacement during the stance phase. In contrast to our original hypothesis we did not observe a relationship between hamstring and gastrocnemius muscle activation and knee flexion angle or displacement.

Anatomically, the quadriceps serve to extend the knee while the gluteus maximus serves to extend the hip. While we only examined knee flexion angle, it is likely that individuals landing with a small knee flexion angle landed in a more erect posture. Previous studies have shown that subjects landing with small knee flexion angles also landed with small hip flexion angles (Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). We hypothesize that individuals landing with small knee flexion angles and a decreased amount of displacement selectively utilized greater quadriceps and gluteus maximus activity to facilitate this more erect body posture and motion during landing. While increased quadriceps and gluteus maximus activity may help to facilitate landing with a small knee flexion angle, these activation strategies in a less flexed body posture may have negative consequences on the ACL.

We chose to investigate knee flexion angle during a jump-landing task because of its close relationship with ACL loading. Sell et al (Sell, Ferris et al. 2007) reported that knee flexion angle during a vertical stop-jump task was significantly and strongly correlated with proximal anterior tibial shear force (PATSF). As discussed earlier, PATSF has been shown in several studies to be the primary ACL loading mechanism (Markolf, Gorek et al. 1990; Markolf, Burchfield et al. 1995). Fleming et al (Fleming, Renstrom et al. 2001) reported that at a fixed knee flexion angle of 20° in vivo ACL strain significantly increased as PASTF increased. Since knee flexion has been shown to be strongly related to PATSF, an understanding of factors influencing knee flexion may help us to gain a better understanding of ACL loading and injury mechanisms.

The PATSF produced by the quadriceps that results in ACL loading is dependent on the patella tendon-tibia shaft angle, the angle between the patella tendon and the

longitudinal axis of the tibia. At a constant level of quadriceps force, the larger the patella tendon-tibia shaft angle, the larger the PATSF (Withrow, Huston et al. 2006; Yu and Garrett 2007). Patella tendon-tibia shaft angle is a function of knee flexion angle as decreased amounts of knee flexion angle result in a greater patella tendon-tibia shaft angle (Nunley 2003). Therefore, the patella tendon-tibia shaft angle influences the amount of PATSF generated by the quadriceps as larger patella tendon-tibia shaft angles result in higher quadriceps induced PATSF and ACL loading. Recent cadaveric studies have reported that ACL relative strain was highly correlated with an increase in quadriceps force (Withrow, Huston et al. 2006). Our findings show that small knee flexion angles are associated with high levels of quadriceps activation. Individuals landing with small knee flexion angles may experience higher PATSF and ACL loading as they increase the patella tendon-tibia shaft angle and utilize greater quadriceps activation. Thus, based on the relationships found in this study, we speculate that landing with small amounts of knee flexion may be a risk factor for non-contact ACL injury.

The lack of significant correlation between knee flexion angles and hamstring activation may also have important clinical implications. This finding suggests that individuals do not appropriately scale the activation amplitude of the hamstrings when increasing quadriceps activation to decrease knee flexion angle and displacement. As individuals land with less knee flexion our study shows that they also increase quadriceps and gluteus maximus activation. However, the hamstring activation was not increased in similar magnitudes and no significant correlation was revealed between hamstrings activation with knee flexion angle and displacement. The lack of increase in hamstring and gastrocnemius activation in combination with increased quadriceps activation

suggests that as individuals land with small knee flexion angles they are utilizing less co-activation of the knee extensors (quadriceps) and flexors (hamstrings and gastrocnemius), displaying a quadriceps dominant strategy. The hamstrings have been shown to offset PATSF at the knee, hence minimizing ACL loading (Li, Rudy et al. 1999). Increased quadriceps activity without an increase in hamstring activity may allow for greater PATSF, and possibly ACL loading to occur at the knee.

Based on our findings as quadriceps and gluteus maximus muscle activity increased knee flexion angle decreased. These findings are in agreement with those of previous studies that were able to demonstrate that individuals landing with decreased knee flexion angle showed increased quadriceps activity (Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). However, these studies did not specifically explore the relationship between knee flexion angle and muscle activation patterns. Sell et al (Sell, Ferris et al. 2007) demonstrated that muscle activity of the quadriceps, along with knee flexion angle, peak posterior ground reaction forces and knee flexion/extension moment significantly predicted PASTF. Combined with the negative relationship found in this study, it may be concluded that quadriceps activity and knee flexion angle contribute to PASTF.

Previous studies have shown females landing with decreased hip flexion angles also had decreased knee flexion angles and increased quadriceps activity (Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). The findings of this investigation demonstrate that gluteus maximus muscle activity has a negative relationship with knee flexion angle and combined with quadriceps activity predicts knee flexion angle. Because of the anatomical attachment at the hip, gluteus maximus contraction may predispose an

individual to an extended hip posture. Increased knee extension during landing has been established as a potential mechanism for ACL injury (Colby, Francisco et al. 2000; Sell, Ferris et al. 2007; Yu and Garrett 2007). However, a relationship between muscle activity at the hip and potential injury at the knee has not be established. The results of the current study show that muscle activity at the hip has a relationship with knee motion and influences knee flexion angle during landing.

A recent meta-analysis of neuromuscular intervention programs reported that ACL injury was reduced following an ACL injury prevention program (Hewett, Ford et al. 2006). The analysis revealed that total ACL injuries were reduced in the training group (n=29) versus the control group (n=110) (Hewett, Ford et al. 2006). The authors hypothesize that neuromuscular training likely alters active knee joint stabilization, which may lead to injury prevention (Hewett, Ford et al. 2006). Included in the training programs was feedback to the athletes suggesting they land in a more bent knee position (Hewett, Ford et al. 2006). The results of the current study also suggest a relationship between knee flexion angle and knee and hip extensor muscle activity. Considering the hypothesis that ACL injury may be prevented through a training program and the negative relationship found in the current study, training athletes to land in a more hip and knee flexed position should be included in ACL prevention programs.

Knee and hip flexion angles are easily identified by clinicians in the clinic and on the field. By including a jump-landing task into a pre-season screening for athletes, athletes at risk for ACL injuries may be identified. Once identified, these athletes can begin a neuromuscular training program to learn to land with a more bent knee. These training programs are easy to incorporate and are cost-effective. With the key to treating

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ACL injury being preventing the injury in the first place, it may be concluded that pre-season screenings to identify at risk athletes should be included in every athletic training room.

It is important to consider that muscle forces were not measured in this study, but rather muscle activity levels. Muscle activation levels have been shown to be a determinant of muscle contraction forces (Clancy, Bouchard et al. 2001). Increased quadriceps activity levels may lead to increased quadriceps force which would in turn influence the loads transmitted to the ACL. Withrow et al (Withrow, Huston et al. 2006) were able to demonstrate that the quadriceps alone generated sufficient forces to injure the ACL at small knee flexion angles.

We acknowledge that the current study has several limitations. First, the subjects included were recreationally active people, not necessarily athletes. The conclusions drawn from this study can only be applicable to people who are recreationally active. Also, the jump-landing task was performed in a laboratory setting with various amounts of equipment attached to the subjects. While the task is similar to athletic movement, it is not as unanticipated and subconscious as it would be on the field or court. Careful attention was given to the EMG sensors and reflective markers when they were placed on the subjects, but it is possible the sensors and markers may have moved and produced errors in their recordings. Also, EMG activity levels are not a measurement of muscle forces being produced. It is possible to relate electrical activity within a muscle to potential force creation, but it can not be accurately or definitively used as a measure of muscle forces. Finally, the correlations and regressions run in the current study are only

mathematical analysis of the relationships between EMG activity and knee flexion angles and cannot be used as a true determinant of their relationship.

Future studies are warranted to determine more relationships than those concluded from our study. Further studies investigating the relationship between knee flexion angle and PASTF would be helpful in determining if a true relationship exists between knee flexion angle and PASTF. Considering the relationships found in this study, an investigation into muscle activity following a neuromuscular training program is warranted in order to determine if training can alter muscle activity and potentially prevent ACL injury.

The current study investigated the relationship between lower extremity muscle activity as measured by EMG and knee flexion angle. Based on the results of this study it can be concluded:

1. Increased quadriceps and gluteus maximus activation, along with a lack of hamstring and gastrocnemius activation, showed a relationship with decreased knee flexion angle and we speculate that this movement pattern may be a potential risk-factor for ACL injury.

2. While it appears that sagittal plane biomechanics and muscle activity have relationships with knee movement, because all of the variance was not explained by these factors, future research is needed to determine if other factors in the transverse and frontal planes influence knee motion.

Table 1: Descriptive Statistics: Demographics (Mean ± SD)

	N	Age (years)	Height (cm)	Mass (kg)
Male	15	22.2 ± 1.78	183.36 ± 6.92	82.21 ± 11.91
Female	15	21.07 ± 2.12	164.53 ± 7.38	62.93 ± 8.91
Total	30	21.63 ± 2.01	173.95 ± 11.88	72.57 ± 14.25

Table 2: Descriptive Statistics: EMG activity normalized to MVIC

Muscle	Phase	N	Mean ± SD
VMO	Preact	29	0.502 ± 0.33
	Decel	29	2.248 ± 0.85
VL	Preact	30	0.282 ± 0.19
	Decel	30	2.48 ± 1.34
GMAX	Preact	29	0.148 ± 0.11
	Decel	29	1.365 ± 1.15
BF	Preact	29	0.151 ± 0.1
	Decel	29	1.156 ± 0.91
LG	Preact	30	0.3 ± 0.18
	Decel	30	1.047 ± 0.56
Quad:Ham ratio	Preact	29	3.49 ± 2.55
	Decel	29	3.04 ± 2.08

Table 3: Descriptive Statistics: Knee flexion angle (Mean ± SD)

	N	Knee flexion angle at IC	Peak knee flexion angle	Average knee flexion displacement
Total	29	20.42°±8.6°	87.42°±18.29°	66.99°±17.79°

Table 4: Correlations: Muscle preactivation and knee flexion at IC

Muscle and Phase		Average Knee Flexion at IC
VMO preact	Pearson Correlation	-.382*
	Significance (2-tailed)	.045
	N	28
VL preact	Pearson Correlation	-.190
	Significance (2-tailed)	.322
	N	29
GMAX preact	Pearson Correlation	-.385*
	Significance (2-tailed)	.043
	N	28
BF preact	Pearson Correlation	.214
	Significance (2-tailed)	.274
	N	28
LG preact	Pearson Correlation	-.313
	Significance (2-tailed)	.098
	N	29
Quad:Ham ratio	Pearson Correlation	-.442*
	Significance (2-tailed)	.018
	N	28

*Correlation is significant at the 0.05 level

Table 5: Correlations: Muscle activity during deceleration and peak knee flexion

Muscle and Phase		Peak Knee Flexion Angle
VMO decel	Pearson Correlation	-.687**
	Significance (2-tailed)	.000
	N	28
VL decel	Pearson Correlation	-.467*
	Significance (2-tailed)	.011
	N	29
GMAX decel	Pearson Correlation	-.386*
	Significance (2-tailed)	.043
	N	28
BF decel	Pearson Correlation	-.114
	Significance (2-tailed)	.562
	N	28
LG decel	Pearson Correlation	-.160
	Significance (2-tailed)	.406
	N	29
Quad:Ham ratio	Pearson Correlation	-.369
	Significance (2-tailed)	.053
	N	29

*Correlation is significant at the 0.05 level
**Correlation is significant at the 0.01 level

Table 6: Correlations: Muscle activity during deceleration and knee flexion displacement

Muscle and Phase		Average Knee Flexion Displacement
VMO decel	Pearson Correlation	-.631**
	Significance (2-tailed)	.000
	N	28
VL decel	Pearson Correlation	-.453*
	Significance (2-tailed)	.014
	N	29
GMAX decel	Pearson Correlation	-.235
	Significance (2-tailed)	.229
	N	28
BF decel	Pearson Correlation	-.177
	Significance (2-tailed)	.369
	N	28
LG decel	Pearson Correlation	-.190
	Significance (2-tailed)	.323
	N	29
Quad:Ham ratio	Pearson Correlation	-.272
	Significance (2-tailed)	.162
	N	28

*Correlation is significant at the 0.05 level
**Correlation is significant at the 0.01 level

| Predictor Variable(s)	Criterion Variable(s)	R	R²	P	R² Change	F change	Sig. F change
Q:H ratio	Knee flexion IC	.461	.213	.015	.213	6.763	.015
Q:H ratio, GMAX preact	**Knee flexion IC**	**.589**	**.347**	**.006**	**.134**	**4.931**	**.036**
Q:H ratio, GMAX, VMO preact	Knee flexion IC	.591	.349	.018	.002	.085	.773
VMO decel	**Peak knee flexion angle**	**.686**	**.471**	**.000**	**.471**	**23.154**	**.000**
VMO, VL decel	Peak knee flexion angle	.702	.492	.000	.021	1.045	.317
VMO, VL, GMAX decel	Peak knee flexion angle	.714	.509	.001	.017	.837	.369
VMO decel	Average knee flexion displacement	.631	.399	.000	.399	17.241	.000
VMO, VL decel	**Average knee flexion displacement**	**.704**	**.495**	**.000**	**.096**	**4.774**	**.038**

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APPENDIX B: FIGURES

Figure 1: Knee flexion (IC) at Q:H ratio and norm GMAX
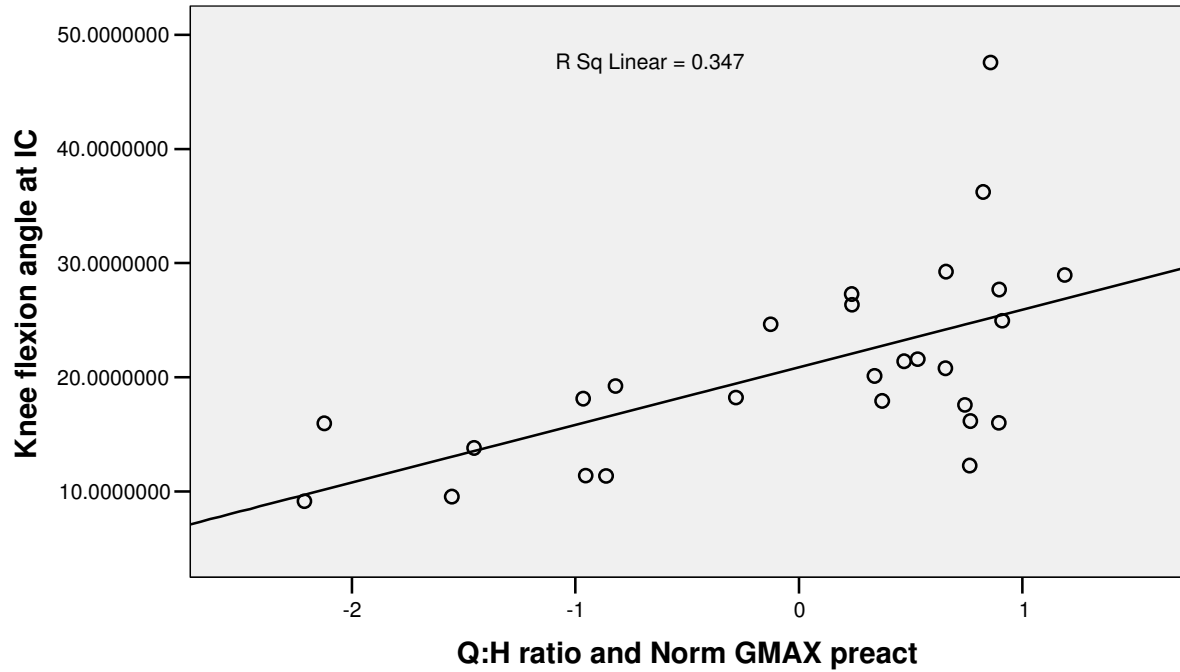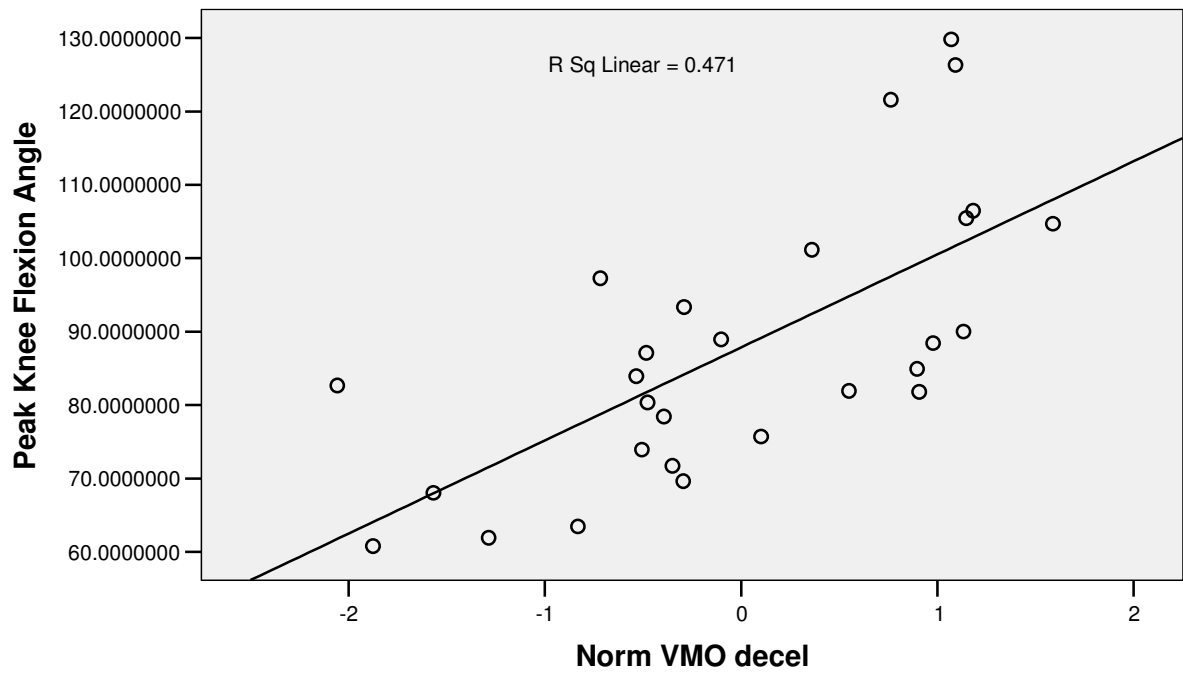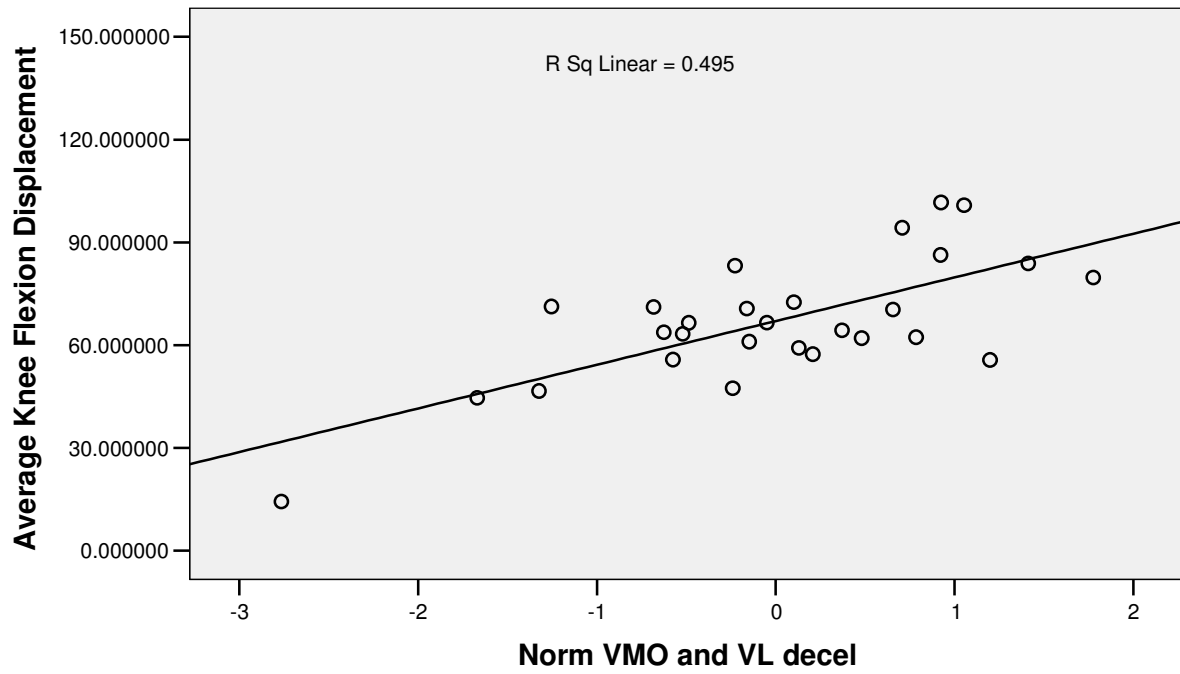
Figure 2: Peak knee flexion angle at norm VMO

Figure 3: Knee flexion displacement at norm VMO and VL decel

Introduction

A milestone for female athletics was the passage of Title XI in 1972, with participation by females in athletics growing immensely since. Along with the increase in participation has come an increase in injuries in female athletes, notably anterior cruciate ligament (ACL) injuries (Arendt, Agel et al. 1999; Agel, Arendt et al. 2005). Researchers have shown that females participating in sports requiring pivoting and jumping suffer ACL injuries at a 4-6 times greater rate than males participating in the same sport (Hewett, Myer et al. 2005). Although ACL injury has become a focus of sports medicine research, the explanation behind the prevalence of females suffering non-contact ACL injury has remained largely unclear. Several risk factors for female non-contact ACL injuries have been suggested, focusing on three main etiologic areas: anatomical, hormonal and neuromuscular (Hewett, Myer et al. 2005). Although anatomical and hormonal differences have been identified between females and males, these factors remain mainly non-modifiable. An increasing amount of evidence suggests that altered muscle activation and joint positioning during high risk deceleration activities such as running, cutting, pivoting, or landing may be the main contributing factors to female non-contact ACL injury (Malinzak, Colby et al. 2001; Hewett, Myer et al. 2005). Muscle activation patterns and joint positions that directly influence ACL loading are believed to play an important role in the elevated risk of ACL injury observed in females.

Several studies have shown females to have different lower extremity motion and muscle activity patterns than their male counterparts during non-contact athletic tasks (Malinzak, Colby et al. 2001; Hewett, Myer et al. 2005; Chappell, Creighton et al. 2007).

Specifically, females have been shown to perform athletic tasks with decreased knee and hip flexion angles, (Malinzak, Colby et al. 2001; Yu, Lin et al. 2006; Chappell, Creighton et al. 2007) as well as a trend towards higher quadriceps, soleus, and gastrocnemius activity and lower hamstring activity compared to males (Malinzak, Colby et al. 2001; White, Lee et al. 2003; Padua, Carcia et al. 2005; Kvist 2006) It is hypothesized that the observed sex differences in muscle activation and knee flexion angle during cutting, jumping, landing, and running may partially explain the sex differences in non-contact ACL injury rates (Hewett, Ford et al. 2006; Yu, Lin et al. 2006; Chappell, Creighton et al. 2007; Sell, Ferris et al. 2007).

Research suggests that the neuromuscular control over the lower extremity may play a crucial role in the mechanism behind ACL injury. Muscle activation strategies coupled with knee flexion angle likely influence strain at the ACL. However, the literature lacks data describing the collective effects of sagittal plane musculature of the lower extremity on the knee during athletic tasks. Therefore, the purpose of our study was to determine relationships between lower extremity muscle activity (quadriceps, hamstrings, gluteus maximus, and gastrocnemius) and knee flexion angle during a jump-landing task.

Research Questions

1. Is there a relationship between quadriceps, hamstring, gluteus maximus, and gastrocnemius EMG amplitude and knee flexion angle during a jump landing task?

2. Does a combination of quadriceps, hamstring, gluteus maximus, and gastrocnemius EMG predict knee flexion angle during a jump landing task?

Research Hypothesis

1. There will be a negative relationship between quadriceps, gluteus maximus, and gastrocnemius EMG amplitude and knee flexion angle during a jump landing task. There will be a positive relationship between hamstring EMG amplitude and knee flexion angle during a jump landing task.

2. The combination of quadriceps, hamstring, gluteus maximus, and gastrocnemius EMG activity will predict knee flexion angle during a jump landing task.

Subjects

Thirty subjects (15 males, 15 females) between the ages of 18-30 years were recruited from The University of North Carolina at Chapel Hill campus. Power analysis for a regression model indicated that for an effect size of 0.8 approximately 30 subjects were necessary. Inclusion criteria included: 1.) participation in physical activity for 30 minutes a day, a minimum of 3 days a week 2.) current (intramural or club) or former (at least one year of high school varsity) participation in organized soccer, volleyball, basketball, or lacrosse. Subjects were free from lower extremity injury in either leg within the past 6 months and were also excluded if they had a history of surgery to the lower extremity within the past 2 years or a history of ACL surgery.

Instrumentation

A surface electromyography (EMG) system (Delsys Bagnoli-8, Boston, MA) was worn by the subject and used to record muscle activity of the gluteus maximus, quadriceps (vastus medialis oblique and vastus lateralis), hamstrings (biceps femoris), and lateral gastrocnemius, via surface electrodes. Raw EMG was collected via Vicon Nexus Software (Vicon Motion Systems, Centennial, CO) and stored for analysis. Unit specifications for the EMG system included a CMRR of 92 dB and amplifier gain of

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1000. A force plate (Bertec Corportation, Columbus, OH) was used to collect kinetic

data. Force plate and EMG data were collected at 1200Hz. Each subject was outfitted

with reflective markers to record kinematic data of the lower extremity during the jump

landing task. The movement of the reflective markers was captured by 7 infrared video

cameras (Vicon Motion Systems, Centennial, CO) at a frame rate of 120 Hz. All data

were collected using Vicon Nexus Software (Vicon Motion Systems, Centennial, CO). A

global reference system was defined using the right hand rule, in which the x-axis is

positive in the anterior direction, the y-axis is positive to the left of each subject, and the

z-axis is positive in the superior direction

Procedures

Subjects reported to the Sports Medicine Research Laboratory at the University of

North Carolina at Chapel Hill for a single testing session lasting approximately 1.5 hours.

Subjects were required to wear athletic shoes, and Lycra spandex shorts and shirts were

provided to each subject. Upon arrival all subjects read and signed a consent form

approved by The University of North Carolina Biomedical Institutional Review Board.

Demographic information was collected for each subject and a health questionnaire was

used to assess lower extremity injury status. Subjects then completed a 5-minute warm-

up on a stationary cycle ergometer at a self-selected pace.

The dominant leg was defined as the leg used to kick a ball for maximum distance

and was used for kinematic and EMG data collection for each subject. For EMG

preparation, each subject's skin was shaved, abraded and cleaned with isopropyl alcohol

prior to application of surface electrodes. The electrodes for the gluteus maximus were

placed over the greatest prominence of the middle of the buttocks and midway between

the sacral vertebrae and the greater trochanter (Basmajian, Blumenstein, and Dismatsek, 1980). The electrodes for the hamstrings were placed over the biceps femoris near the measured midpoint of the muscle belly (Basmajian, Blumenstein, and Dismatsek, 1980). The electrodes for the quadriceps were placed over the vastus lateralis, inferolateral to 50% of the quadriceps length (Basmajian, Blumenstein, and Dismatsek, 1980). The electrodes for the gastrocnemius were placed over the bulge of the lateral head of the gastrocnemius (Basmajian, Blumenstein, and Dismatsek, 1980). Electrode placement was confirmed with manual muscle testing. Subjects were then outfitted with reflective markers placed on the following landmarks: Right and left acromion processes, right and left anterior superior iliac spines, S1 joint space, right and left greater trochanters, lateral aspects of the right and left thighs, lateral epicondyle of the right and left knee, medial epicondyle of the right and left knee, lateral aspects of the right and left shanks, right and left lateral malleoli, right and left medial malleoli, right and left heels, the heads of the right and left 5th metatarsals, and the head of the right and left 1st metatarsals. The markers were affixed to the skin and shoes with adhesive tape. Following marker placement, the participant was asked to stand in the center of the calibration area (2.5 m high × 2.5 m long × 1.5 m wide) with each foot on a Bertec Forceplate (Type 4060-08, Bertec Corporation, Worthington, OH), in order to collect a static calibration trial. Following the static calibration trial the medial malleolus and epicondyle markers were removed for data collection during the jump landing task.

EMG activity of the gluteus maximus, quadriceps, hamstrings, and gastrocnemius, and kinematic data were collected during 10 trials of a jump landing task. The task was first described then demonstrated to the subject. For the task, a 30cm box

was placed a distance of 50% of subject's height from the force plate. Subjects were instructed to jump down from the box directly onto the force plates, landing with one foot in the middle of each force plate, then jump straight up for maximum height and land with both feet back on each force plate. The jump-landing task was similar to those previously investigated (Hewett, Myer et al. 2005). Subjects were allowed a maximum of 5 practice trials prior to data collection. A one minute rest period between each of the test trials was given to each subject to avoid fatigue.

Following the data collection trials, each subject performed three maximum voluntary isometric contractions (MVIC) for the gluteus maximus, quadriceps, hamstrings, and gastrocnemius. Each trial was held for 5 seconds. For the gluteus maximus MVIC, the subject was positioned on a table prone, knee flexed to 90 degrees, with a strap placed over the mid-belly of the hamstrings. Subjects were instructed to contract isometrically into hip extension. For MVIC of the quadriceps and hamstrings, subjects were positioned sitting with hips and knees at 90 degrees in a Kincom chair with straps around their legs and trunk. For quadriceps MVIC, subjects were instructed to kick into the strap, extending at the knee. For hamstrings MVIC, subjects were instructed to kick back into the strap, flexing the knee. For gastrocnemius MVIC, subjects laid on the table prone with knees fully extended. A strap was placed around the metatarsal heads of the subject and the subjects were instructed to push into the strap with maximum force. These testing positions are similar to the manual muscle tests described by Hislop and Montgomery (Hislop, Montgomery, 2002).

Data Reduction

Peak knee flexion angle and vertical ground reaction force were used to define the phases of the jump-landing task. Initial contact was defined as the time point in which the vertical ground reaction force exceeded 10N. The preactivation phase was defined as 200ms prior to initial contact. The deceleration phase was defined as the phase from initial contact to the point of peak knee flexion. The stance phase was defined as the phase from initial contact to takeoff. Takeoff was defined as the time point in which vertical ground reaction force was less than 10N.

All EMG data were passively demeaned, bandpass filtered between 10-350Hz, notch filtered between 59.5-60.5Hz, and smoothed using a 20ms root mean squared sliding window function. Mean EMG amplitude of the gluteus maximus, quadriceps, hamstrings, and gastrocnemius were calculated separately for each of the jump-landing trials for the preactivation, deceleration, and stance phases. All EMG data was normalized to the maximum voluntary isometric contraction (MVIC) for each subject during each phase. The mean EMG amplitude was determined during the middle three seconds of each of the five second MVIC trials and the trials were averaged. Normalized EMG was used for data analysis. The quadriceps:hamstring ratio (Q:H ratio) was calculated for each of the phases by calculating the arithmetic mean of the normalized VMO and VL EMG data and then dividing by the normalized hamstrings EMG data.

Knee flexion angle at initial contact was calculated for each subject for each of the ten trials of the jump-landing task. Peak knee flexion angle and knee flexion displacement during the deceleration and stance phases of the jump-landing task were calculated for each subject for each of the ten trials. Knee flexion displacement during the deceleration and stance phases were determined by subtracting the knee flexion angle

at initial contact from the peak knee flexion angle during each phase. The arithmetic mean of knee flexion angle at initial contact was calculated for each subject across the ten trials of the jump-landing task. The arithmetic mean of peak knee flexion angle and displacement were calculated during the deceleration and stance phases across the ten trials of the jump-landing task. Mean knee flexion values were used for statistical analysis.

Statistical Analysis

All data analyses were performed using SPSS version 13.0 (SPSS, Inc. Chicago, IL). Kinematic and EMG data were averaged across the 10 trials for each participant. The reduced kinematic and EMG data collected for each subject were analyzed using correlation and regression analysis to determine relationships between EMG data and knee flexion angle and displacement during each phase of the jump-landing task. Individual simple correlational analyses were run for the relationship between knee flexion angle and displacement and EMG mean amplitude for the gluteus maximus, quadriceps, hamstrings, and gastrocnemius during each phase of the jump-landing task. The variables that were found to have significant correlations were then entered into a forward stepwise multiple linear regression analysis used to predict knee flexion angle and displacement during each phase of the jump-landing task. Variables entered into the regression models were entered in order of highest significance. A priori alpha level was set at 0.05.

Results

Correlation Analysis

Fifteen male (age= 22.2 ± 1.78 years, height = 183.36 ± 6.92 cm, mass = 82.21 ±

11.91 kg) and fifteen female subjects (age= 21.07 ± 2.12 years, height = 164.53 ± 7.38

cm, mass = 62.93 ± 8.91 kg) completed testing for this investigation. Six subjects had

trials that were not included in the statistical analysis due to the EMG data and kinematic

data being outliers (above three standard deviations from the mean). The data that were

not included in the statistical analysis are as follows: subject 1, EMG data of vastus

medialis oblique (VMO) at preactivation and stance phases and all EMG data of the

gluteus maximus; subject 2, all biceps femoris EMG data; subject 9, knee flexion data at

initial contact; and subject 31, EMG data of the VMO during the deceleration phase.

Additionally, subject 4 only had 8 trials of biceps femoris EMG data during the

deceleration phase that were analyzed due to the trial data being above three standard

deviations from the mean; subject 27 had only 9 trials of vastus lateralis EMG data

during both the preactivation and stance phase analyzed due to the trial data being above

three standard deviations from the mean.

Eight significant correlations were found ($P<0.05$) (Tables 4-6). During the

preactivation phase the EMG activity of the VMO and the gluteus maximus (GMAX) and

the Q:H ratio were found to have significant negative relationships with knee flexion

angle at initial contact (VMO: r = -0.382, P = 0.045, GMAX: r = -0.385, P = 0.043 ,Q:H

ratio: r = -.442, P = .018). During the deceleration phase EMG activity of the VMO,

vastus lateralis (VL) and GMAX were found to have significant negative relationships

with peak knee flexion angle during the stance phase (VMO: r = -0.687, P =0.00, VL: r =

-0.467, P = .011, GMAX: r = -0.386, P = 0.043). Also, during the deceleration phase

EMG activity of the VMO and VL demonstrated significant negative relationships with

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knee flexion displacement during the stance phase (VMO: r = -0.631, *P*= 0.00, VL: r = -0.453, *P* = 0.014). No significant correlations were found between biceps femoris or lateral gastrocnemius EMG activity during the preactivation or deceleration phases and knee flexion angle at initial contact, peak knee flexion angle or knee flexion displacement during the stance phase (*P* >.05) (Tables 4-6). The negative relationships revealed between knee flexion and various measures of EMG activity indicate that decreased knee flexion angle and displacement were associated with increased EMG activity of the VMO, VL, and GMAX muscles and Q:H ratio.

Regression Analysis

The dependent variables that were found to have significant simple correlations were included in the regression analysis (Table 7). A forward stepwise multiple linear regression analysis was used including R^2 change statistics calculated. The regression models presented are those with the highest R^2 change significance (Table 7). Q:H ratio and EMG activity of the GMAX during the preactivation phase were found to predict approximately 34.7% of the variance in knee flexion angle at initial contact (R^2= 0.347, *P* = 0.006) (Figure 1). During the deceleration phase EMG activity of the VMO was found to predict approximately 47.1% of the variance in peak knee flexion angle during the stance phase (R^2= 0.471, *P* = 0.000) (Figure 2). Finally, EMG activity of the VMO and VL during the deceleration phase were shown to predict approximately 49.5% of the variance in knee flexion displacement during the stance phase (R^2= 0.495, *P* = 0.000) (Figure 3).

Discussion

The purpose of this study was to explore the relationships between muscle activity and knee flexion angle during a jump-landing task in recreationally active males and females. The primary findings of this investigation support a negative relationship between muscle activity of the quadriceps and gluteus maximus musculature and knee flexion angle and displacement during a jump-landing task. In support of our original hypothesis we observed that increased activation of the quadriceps and gluteus maximus musculature was related to a decrease in sagittal plane motion of the knee during a jump-landing task. When assessing the co-activation ratio of quadriceps and hamstrings activity along with gluteus maximus muscle activity during the jump-landing task, we found that preactivation of these muscles predicted knee flexion angle at initial contact. Also, we found that quadriceps muscle activity during the deceleration phase predicted nearly half of the variance in peak knee flexion angle and knee flexion displacement during the stance phase. In contrast to our original hypothesis we did not observe a relationship between hamstring and gastrocnemius muscle activation and knee flexion angle or displacement.

Anatomically, the quadriceps serve to extend the knee while the gluteus maximus serves to extend the hip. While we only examined knee flexion angle, it is likely that individuals landing with a small knee flexion angle landed in a more erect posture. Previous studies have shown that subjects landing with small knee flexion angles also landed with small hip flexion angles (Yu, Lin et al. 2006; Chappell, Creighton et al. 2007). We hypothesize that individuals landing with small knee flexion angles and a decreased amount of displacement selectively utilized greater quadriceps and gluteus maximus activity to facilitate this more erect body posture and motion during landing.

While increased quadriceps and gluteus maximus activity may help to facilitate landing with a small knee flexion angle, these activation strategies in a less flexed body posture may have negative consequences on the ACL.

We chose to investigate knee flexion angle during a jump-landing task because of its close relationship with ACL loading. Sell et al (Sell, Ferris et al. 2007) reported that knee flexion angle during a vertical stop-jump task was significantly and strongly correlated with proximal anterior tibial shear force (PATSF). As discussed earlier, PATSF has been shown in several studies to be the primary ACL loading mechanism (Markolf, Gorek et al. 1990; Markolf, Burchfield et al. 1995). Fleming et al (Fleming, Renstrom et al. 2001) reported that at a fixed knee flexion angle of 20° in vivo ACL strain significantly increased as PASTF increased. Since knee flexion has been shown to be strongly related to PATSF, an understanding of factors influencing knee flexion may help us to gain a better understanding of ACL loading and injury mechanisms.

The PATSF produced by the quadriceps that results in ACL loading is dependent on the patella tendon-tibia shaft angle, the angle between the patella tendon and the longitudinal axis of the tibia. At a constant level of quadriceps force, the larger the patella tendon-tibia shaft angle, the larger the PATSF (Withrow, Huston et al. 2006; Yu and Garrett 2007). Patella tendon-tibia shaft angle is a function of knee flexion angle as decreased amounts of knee flexion angle result in a greater patella tendon-tibia shaft angle (Nunley 2003). Therefore, the patella tendon-tibia shaft angle influences the amount of PATSF generated by the quadriceps as larger patella tendon-tibia shaft angles result in higher quadriceps induced PATSF and ACL loading. Recent cadaveric studies have reported that ACL relative strain was highly correlated with an increase in

quadriceps force (Withrow, Huston et al. 2006). Our findings show that small knee flexion angles are associated with high levels of quadriceps activation. Individuals landing with small knee flexion angles may experience higher PATSF and ACL loading as they increase the patella tendon-tibia shaft angle and utilize greater quadriceps activation. Thus, based on the relationships found in this study, we speculate that landing with small amounts of knee flexion may be a risk factor for non-contact ACL injury.

The lack of significant correlation between knee flexion angles and hamstring activation may also have important clinical implications. This finding suggests that individuals do not appropriately scale the activation amplitude of the hamstrings when increasing quadriceps activation to decrease knee flexion angle and displacement. As individuals land with less knee flexion our study shows that they also increase quadriceps and gluteus maximus activation. However, the hamstring activation was not increased in similar magnitudes and no significant correlation was revealed between hamstrings activation with knee flexion angle and displacement. The lack of increase in hamstring and gastrocnemius activation in combination with increased quadriceps activation suggests that as individuals land with small knee flexion angles they are utilizing less co-activation of the knee extensors (quadriceps) and flexors (hamstrings and gastrocnemius), displaying a quadriceps dominant strategy. The hamstrings have been shown to offset PATSF at the knee, hence minimizing ACL loading (Li, Rudy et al. 1999). Increased quadriceps activity without an increase in hamstring activity may allow for greater PATSF, and possibly ACL loading to occur at the knee.

Based on our findings as quadriceps and gluteus maximus muscle activity increased knee flexion angle decreased. These findings are in agreement with those of

previous studies that were able to demonstrate that individuals landing with decreased

knee flexion angle showed increased quadriceps activity (Yu, Lin et al. 2006; Chappell,

Creighton et al. 2007). However, these studies did not specifically explore the

relationship between knee flexion angle and muscle activation patterns. Sell et al (Sell,

Ferris et al. 2007) demonstrated that muscle activity of the quadriceps, along with knee

flexion angle, peak posterior ground reaction forces and knee flexion/extension moment

significantly predicted PASTF. Combined with the negative relationship found in this

study, it may be concluded that quadriceps activity and knee flexion angle contribute to

PASTF.

Previous studies have shown females landing with decreased hip flexion angles

also had decreased knee flexion angles and increased quadriceps activity (Yu, Lin et al.

2006; Chappell, Creighton et al. 2007). The findings of this investigation demonstrate

that gluteus maximus muscle activity has a negative relationship with knee flexion angle

and combined with quadriceps activity predicts knee flexion angle. Because of the

anatomical attachment at the hip, gluteus maximus contraction may predispose an

individual to an extended hip posture. Increased knee extension during landing has been

established as a potential mechanism for ACL injury (Colby, Francisco et al. 2000; Sell,

Ferris et al. 2007; Yu and Garrett 2007). However, a relationship between muscle

activity at the hip and potential injury at the knee has not be established. The results of

the current study show that muscle activity at the hip has a relationship with knee motion

and influences knee flexion angle during landing.

A recent meta-analysis of neuromuscular intervention programs reported that

ACL injury was reduced following an ACL injury prevention program (Hewett, Ford et

al. 2006). The analysis revealed that total ACL injuries were reduced in the training group (n=29) versus the control group (n=110) (Hewett, Ford et al. 2006). The authors hypothesize that neuromuscular training likely alters active knee joint stabilization, which may lead to injury prevention (Hewett, Ford et al. 2006). Included in the training programs was feedback to the athletes suggesting they land in a more bent knee position (Hewett, Ford et al. 2006). The results of the current study also suggest a relationship between knee flexion angle and knee and hip extensor muscle activity. Considering the hypothesis that ACL injury may be prevented through a training program and the negative relationship found in the current study, training athletes to land in a more hip and knee flexed position should be included in ACL prevention programs.

Knee and hip flexion angles are easily identified by clinicians in the clinic and on the field. By including a jump-landing task into a pre-season screening for athletes, athletes at risk for ACL injuries may be identified. Once identified, these athletes can begin a neuromuscular training program to learn to land with a more bent knee. These training programs are easy to incorporate and are cost-effective. With the key to treating ACL injury being preventing the injury in the first place, it may be concluded that pre-season screenings to identify at risk athletes should be included in every athletic training room.

It is important to consider that muscle forces were not measured in this study, but rather muscle activity levels. Muscle activation levels have been shown to be a determinant of muscle contraction forces (Clancy, Bouchard et al. 2001). Increased quadriceps activity levels may lead to increased quadriceps force which would in turn influence the loads transmitted to the ACL. Withrow et al (Withrow, Huston et al. 2006)

were able to demonstrate that the quadriceps alone generated sufficient forces to injure the ACL at small knee flexion angles.

We acknowledge that the current study has several limitations. First, the subjects included were recreationally active people, not necessarily athletes. The conclusions drawn from this study can only be applicable to people who are recreationally active. Also, the jump-landing task was performed in a laboratory setting with various amounts of equipment attached to the subjects. While the task is similar to athletic movement, it is not as unanticipated and subconscious as it would be on the field or court. Careful attention was given to the EMG sensors and reflective markers when they were placed on the subjects, but it is possible the sensors and markers may have moved and produced errors in their recordings. Also, EMG activity levels are not a measurement of muscle forces being produced. It is possible to relate electrical activity within a muscle to potential force creation, but it can not be accurately or definitively used as a measure of muscle forces. Finally, the correlations and regressions run in the current study are only mathematical analysis of the relationships between EMG activity and knee flexion angles and cannot be used as a true determinant of their relationship.

Future studies are warranted to determine more relationships than those concluded from our study. Further studies investigating the relationship between knee flexion angle and PASTF would be helpful in determining if a true relationship exists between knee flexion angle and PASTF. Considering the relationships found in this study, an investigation into muscle activity following a neuromuscular training program is warranted in order to determine if training can alter muscle activity and potentially prevent ACL injury.

The current study investigated the relationship between lower extremity muscle activity as measured by EMG and knee flexion angle. Based on the results of this study it can be concluded:

1. Increased quadriceps and gluteus maximus activation, along with a lack of hamstring and gastrocnemius activation, showed a relationship with decreased knee flexion angle and we speculate that this movement pattern may be a potential risk-factor for ACL injury.

2. While it appears that sagittal plane biomechanics and muscle activity have relationships with knee movement, because all of the variance was not explained by these factors, future research is needed to determine if other factors in the transverse and frontal planes influence knee motion.

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