Group comparison of lower extremity muscle activation and lower extremity muscular flexibility and their effect on single leg squat performance

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ABSTRACT

Timothy C. Mauntel: Group comparison of lower extremity muscle activation and lower extremity muscular flexibility and their effect on single leg squat performance. (Under the direction of Darin Padua, Troy Blackburn, and Rebecca Begalle).

Knee valgus is a potential risk factor for lower extremity (LE) injuries. LE movement screenings and flexibility measurements may be utilized to identify neuromuscular patterns, which contribute to knee valgus. There are few studies that have investigated how flexibility and muscular activation differ between individuals who display medial knee displacement (MKD) during a single leg squat (SLS) and those who do not. We hypothesized that flexibility and muscular activation would differ between the groups. Forty individuals completed flexibility measurements and a SLS task while EMG data were collected from eight LE muscles. Three MANOVAs were run comparing flexibility measurements, EMG data, and muscle co-activation ratios. The MKD group had significantly less dorsiflexion, greater talar glide motion, and smaller hip adductor and gluteal co-activation ratios compared to the control group. Therefore, knee valgus appears to be influenced by decreased dorsiflexion and decreased co-activation of the Hip ADD and GMed and GMax muscles.

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CHAPTER I

INTRODUCTION

BACKGROUND

Noncontact knee injuries are a common occurrence among physically active populations. In particular, physically active people are susceptible to injury of the anterior cruciate ligament (ACL) (Noyes, Mooar et al. 1983; Hewett, Lindenfeld et al. 1999; Griffin, Albohm et al. 2006), and medial collateral ligament (MCL) (Fetto and Marshall 1978; Griffith, LaPrade et al. 2009; Wijdicks, Griffith et al. 2009), and to developing patellofemoral pain syndrome (PFPS) (Devereaux and Lachmann 1984; Stathopulu and Baildam 2003). There are an estimated 80,000 to 250,000 ACL injuries annually; (Griffin, Albohm et al. 2006) approximately 70% of these injuries result from noncontact mechanisms, the majority of which involve landing from a jump (Boden, Griffin et al. 2000; Hewett, Myer et al. 2006). The average cost of surgical repair and rehabilitation is approximately $17,000 per incident (Hewett, Lindenfeld et al. 1999). The MCL is one of the most commonly injured ligaments in the knee (Grood, Noyes et al. 1981; LaPrade 1999). Cases of PFPS also account for a large percentage of all knee injuries seen in sports medicine clinics (up to 25%) (Devereaux and Lachmann 1984) and result in a restriction of physical activity in 36% of all people affected by it (Stathopulu and Baildam 2003). Active persons between the ages of 15 and 25 years are the most commonly afflicted with ACL injury or PFPS (Devereaux and Lachmann 1984; Griffin, Albohm et al. 2006). These injuries not only affect an individual's physical health, but also the financial cost resulting from the treatment

of these injuries, the potential loss of time from competition, and/or the possible loss of scholarship funding can negatively affect a person's mental health (Freedman, Glasgow et al. 1998). These factors make it important to study noncontact knee injuries in order to obtain a better understanding of predisposing factors that might be used to both identify and treat individuals at risk for these injuries. Therefore it is important to develop clinical screening instruments to identify these high risk individuals and provide information to guide rehabilitation strategies.

Malalignment of lower extremity segments during functional activities has been shown to increase the likelihood of sustaining a noncontact knee injury (Hewett, Stroupe et al. 1996; Boden, Griffin et al. 2000; Griffin, Agel et al. 2000; Griffin, Albohm et al. 2006; Hewett, Ford et al. 2006). One of the most common predisposing factors for noncontact knee injuries is greater knee valgus movement during functional tasks, such as landing or cutting (Hughston, Andrews et al. 1976; Hewett, Stroupe et al. 1996; Ford, Myer et al. 2003; Hewett, Myer et al. 2005). It has been suggested that as the knee goes into a valgus or varus position during activity it places the joint in a less stable position and makes it more susceptible to injury (Hewett, Stroupe et al. 1996). Injuries commonly associated with greater dynamic knee valgus movement include injury to the ACL (Kennedy, Weinberg et al. 1974; Hewett, Lindenfeld et al. 1999; Boden, Griffin et al. 2000; Hewett, Myer et al. 2005; Hewett, Ford et al. 2006) and MCL (Hughston, Andrews et al. 1976), and PFPS (Reikeras 1992; Ireland, Willson et al. 2003; Lee, Morris et al. 2003; Powers 2003; Willson and Davis 2008). The association between knee valgus and noncontact knee injuries makes it important to understand those factors which contribute to this movement pattern.

In an effort to combat the high incidence of noncontact knee injuries, especially injuries to the ACL, a number of lower extremity screenings have been developed to aid in identifying risk factors associated with these injuries (Bonci 1999; Hewett, Myer et al. 2001; Crill, Kolba et al. 2004; DiMattia, Livengood et al. 2005; McLean, Walker et al. 2005; Newton, Gerber et al. 2006; Hirth 2007; Willson and Davis 2008; Padua, Marshall et al. 2009). Clinical assessments of those persons who demonstrate dynamic valgus as it relates to injury risk have typically been described during squatting and landing tasks. The Landing Error Scoring System (LESS) (Padua, Marshall et al. 2009), observation of a drop landing task (Ekegren, Miller et al. 2009), the forward lunge (Crill, Kolba et al. 2004), and the overhead squat test (Hirth 2007) are all validated lower extremity screenings which require no equipment and can be effectively used by new clinicians with only minimal instruction (Hirth 2007; Padua, Marshall et al. 2009). The benefits of these screenings are that they are practical, cost effective, and require minimal space to complete. Similar to other clinical screenings the single leg squat has been utilized to identify persons with faulty lower extremity biomechanics, primarily through the observation of knee valgus displacement. Traditionally, greater knee valgus observed during the single leg squat has been attributed to poor hip strength or muscle imbalances and poor neuromuscular control of key hip and trunk musculature (Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Willson, Ireland et al. 2006).

Lower extremity musculature has been suggested as a contributing factor to knee valgus. Many studies have examined the musculature surrounding the hip and its effect on knee valgus movement during a single leg squat (SLS) (Zeller, McCrory et al. 2003; Lawrence, Kernozek et al. 2008). Traditionally, a lot of attention has been given to the

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strength of the hip abductor group, specifically the gluteus medius, and its effect on frontal plane knee movement; however, recent studies have shown little to no correlation between the two (DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006). Therefore, other factors such as lower extremity muscular activation and lower extremity flexibility may play an important role in dynamic valgus.

The activation of the musculature surrounding the hip and knee appears to play a major role in the kinematics of the knee joint during functional activities. A relationship has been established between lesser gluteus medius activation and greater knee valgus movement during a single leg squat (Zeller, McCrory et al. 2003). The musculature acting on the knee has been shown to effect knee frontal plane motion. Decreased neuromuscular control of the rectus femoris (DiMattia, Livengood et al. 2005) and/or increased activation of it during single leg activities has been seen in individuals with greater dynamic knee valgus (Zeller, McCrory et al. 2003). Also, an imbalance in the hamstring-to-quadriceps peak torque ratio (Hewett, Stroupe et al. 1996; Hewett, Myer et al. 2005; Hewett, Myer et al. 2006), and decreased medial-to-lateral firing ratio of the quadriceps and hamstrings during a landing or cutting task can increase knee valgus by compressing the lateral side of the joint while distracting the medial joint space (Lloyd and Buchanan 2001; Ford, Myer et al. 2003; Myer, Ford et al. 2005). Palmieri-Smith et al. reported greater peak knee valgus angles during a single leg forward-jump task in persons with greater activity in the vastus lateralis and biceps femoris, and lesser peak knee valgus angles in those persons with heightened vastus medialis activity (Palmieri-Smith, Wojtys et al. 2008). Decreased quadriceps and hamstring cocontraction activity during a single leg forward-jump task has also been suggested to decrease knee joint stability and, in turn, increase peak knee valgus moments (Palmieri-

Smith, McLean et al. 2009). The medial gastrocnemius acts as a dynamic stabilizer of the knee and helps offset knee valgus moment; therefore decreased activation of the medial gastrocnemius during functional tasks may result in decreased frontal plane stability and result in greater knee valgus (Lloyd and Buchanan 2001). Decreased activation of the medial grastrocnemius during functional tasks has been shown to have significant effect on valgus-varus stability (Lloyd and Buchanan 2001). No previous research has examined the relationships between the activation of the gluteals, hip adductors, hamstrings, quadriceps, and medial head of the gastrocnemius and their influence on lower extremity kinematics during a single leg squat task.

Lower extremity muscular flexibility has also been associated with greater knee instability and dysfunctional kinematics. Greater flexibility of the hamstrings resulting in lesser activation has been suggested to decrease dynamic knee stiffness (Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003) and increase the likelihood of greater knee valgus. One study that examined the causes of PFPS showed that increasing the flexibility of the iliotibial band and hip flexors decreased the symptoms of the participants in the study. One hypothesized explanation for this is that the factors that contribute to tight iliotibial band and hip flexors may cause the pelvis to tilt anteriorly, resulting in femoral internal rotation and therefore altering patellar alignment in the femoral groove (Tyler, Nicholas et al. 2006). It has also been shown that increasing iliotibial band flexibility reduces the pull of the lateral patellar retinaculum and allows the patella to track properly during single-legged partial squat (Powers, Ward et al. 2003). It is has been suggested that tightness of the hip adduction and internal rotation musculature may result in hip adduction and internal rotation motion (Clark and Lucett 2004; Hirth 2007). Similarly if there is increased flexibility of the hip

external rotators this may allow for greater hip internal rotation and further contribute to greater knee valgus during dynamic tasks. Tightness of the lower-leg musculature, especially the lateral gastrocnemius and peroneals, may contribute to tibial abduction and external rotation thus increasing knee valgus (Clark and Lucett 2004; Hirth 2007). It has been suggested that decreased posterior talar mobility would result in decreased dorsi flexion and therefore should be assessed when assessing plantar flexor flexibility (Denegar, Hertel et al. 2002). However, no previous studies have examined the relationships between the flexibility of the hip external rotators, hip internal rotators, hip adductors, hamstrings, iliopsoas, plantar flexors, and talar glide mobility and their effect on lower extremity kinematics during a single leg squat.

Therefore the purpose of this study was to compare lower extremity muscle activation and flexibility between subjects who display medial knee displacement (MKD) and those who do not during a single leg squat. Determining these differences will aid clinicians by identifying the neuromuscular patterns that are associated with different movement patterns that can be discriminated using cost effective clinical assessments and thereby provide a guide for future rehabilitation interventions to correct these faulty mechanics.

VARIABLES

- *Independent*:
    - Group
        - Control: Individuals who maintain a neutral knee position during a single leg squat (Figure 1.1)
        - Medial Knee Displacement (MKD): Individuals who display excessive MKD during a single leg squat (Figure 1.2)

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- *Dependent*:

    - Lower extremity muscle activation intensity (EMG amplitude) during the single leg squat

        - Gluteus maximus
        - Gluteus medius
        - Hip adductors
        - Medial hamstrings
        - Biceps femoris
        - Vastus medialis oblique
        - Vastus lateralis
        - Medial head of gastrocnemius

    - Lower extremity passive range of motion (ROM) muscle flexibility

        - Hip external rotators
        - Hip internal rotators
        - Hip adductors
        - Iliotibial Band
        - Hamstrings (90-90)
        - Hip flexors
        - Plantar flexors (leg straight)
        - Plantar flexors (knee bent)
        - Hip anteversion
        - Posterior talar glide

RESEARCH QUESTIONS & RESEARCH HYPOTHESES

1) *Research Question 1*: Is there a significant difference in lower extremity (LE) muscle activation amplitude between healthy, physically active individuals who demonstrate MKD during a single leg squat compared to those who do not?

    i) *Research Question 1a*: Does mean gluteus maximus EMG amplitude differ between these groups during the descent phase of a single leg squat?

    ii) *Research Question 1b*: Does mean gluteus medius EMG amplitude differ between these groups during the descent phase of a single leg squat?

    iii) *Research Question 1c*: Does mean hip adductor EMG amplitude differ between these groups during the descent phase of a single leg squat?

iv) *Research Question 1d*: Does mean medial hamstrings EMG amplitude differ between these groups during the descent phase of a single leg squat?

v) *Research Question 1e*: Does mean biceps femoris EMG amplitude differ between these groups during the descent phase of a single leg squat?

vi) *Research Question 1f*: Does mean vastus medialis EMG amplitude differ between these groups during the descent phase of a single leg squat?

vii) *Research Question 1g*: Does mean vastus lateralis EMG amplitude differ between these groups during the descent phase of a single leg squat?

viii) *Research Question 1h*: Does mean medial head of the gastrocnemius EMG amplitude differ between these groups during the descent phase of a single leg squat?

- *Hypothesis 1*: The MKD group (MKD Group - MKD) will have one or more LE muscular groups with a significantly higher mean amplitude compared to the control group (Control Group - CG) during a single leg squat.

  i) *Research Hypothesis 1a*: The MKD group will have significantly lower mean amplitude gluteus maximus activation compared to the control group during a single leg squat.

  ii) *Research Hypothesis 1b*: The MKD group will have significantly lower mean amplitude gluteus medius activation compared to the control group during a single leg squat.

  iii) *Research Hypothesis 1c*: The MKD group will have significantly higher mean amplitude hip adductor activation compared to the control group during a single leg squat.

iv) *Research Hypothesis 1d*: The MKD group will have significantly lower mean amplitude medial hamstring activation compared to the control group during a single leg squat.

v) *Research Hypothesis 1e*: The MKD group will have significantly higher mean amplitude biceps femoris activation compared to the control group during a single leg squat.

vi) *Research Hypothesis 1f*: The MKD group will have significantly lower mean amplitude vastus medialis oblique activation compared to the control group during a single leg squat.

vii) *Research Hypothesis 1g*: The MKD group will have significantly higher mean amplitude vastus lateralis activation compared to the control group during a single leg squat.

viii) *Research Hypothesis 1h*: The MKD group will have significantly lower mean amplitude medial head of the gastrocnemius activation compared to the control group during a single leg squat.

2) *Research Question 2*: Is there a significant difference in lower extremity passive range of motion (ROM) muscular flexibility between healthy, physically active individuals who demonstrate excessive knee valgus during a single leg squat compared to those who do not?

i) *Research Question 2a*: Do the hip external rotators mean degrees of passive movement differ between these groups?

ii) *Research Question 2b*: Do the hip internal rotators mean degrees of passive movement differ between these groups?

iii) *Research Question 2c*: Do the hip adductors mean degrees of passive movement differ between these groups?

iv) *Research Question 2d*: Does the iliotibial band mean degrees of passive movement differ between these groups?

v) *Research Question 2e*: Do the hamstrings mean degrees of passive movement differ between these groups?

vi) *Research Question 2f*: Do the hip flexors mean degrees of passive movement differ between these groups?

vii) *Research Question 2g*: Do the plantar flexors (knee extended) mean degrees of passive movement differ between these groups?

viii) *Research Question 2h:* Do the plantar flexors (knee flexed) mean degrees of passive movement differ between these groups?

ix) *Research Question 2i:* Does the hip anteversion mean degrees of femoral neck rotation differ between these groups?

x) *Research Question 2j:* Does the posterior talar glide mean degrees of passive movement differ between these groups?

- *Hypothesis 2*: The MKD group (MKD Group - MKD) will have one or more decreased lower extremity ROM patterns compared to the control group (Control Group - CG).

  i) *Research hypothesis 2a*: The MKD group will have significantly more mean external rotator ROM compared to the control group.

  ii) *Research hypothesis 2b*: The MKD group will have significantly less mean internal rotator ROM compared to the control group.

iii) *Research hypothesis 2c*: The MKD group will have significantly less mean hip abduction ROM compared to the control group.

iv) *Research hypothesis 2d*: The MKD group will have significantly less mean iliotibial ROM compared to the control group.

v) *Research hypothesis 2e*: The MKD group will have significantly more mean hamstring ROM compared to the control group.

vi) *Research hypothesis 2f*: The MKD group will have significantly less mean hip flexor ROM compared to the control group.

vii) *Research hypothesis 2g*: The MKD group will have significantly less mean plantar flexor (knee extended) ROM compared to the control group.

viii) *Research hypothesis 2h*: The MKD group will have significantly less mean plantar flexor (knee flexed) ROM compared to the control group.

ix) *Research hypothesis 2i*: The MKD group will have significantly more hip anteversion mean degrees of femoral neck rotation compared to the control group.

x) *Research hypothesis 2j*: The MKD group will have significantly less posterior talar glide ROM compared to the control group.

Statistical Hypothesis

- Research Question 1

  - Hypothesis 1:

    - H_0: $MKDA = CGA$

    - H_A: $MKDA \neq CGA$

    - H_R: $MKDA > CGA$

- Research Question 2

  - Hypothesis 2:

    - H_0: $_{MKDF} = _{CGF}$

    - H_A: $_{MKDF} \neq _{CGF}$

    - H_R: $_{MKDF} < _{CGF}$

OPERATIONAL DEFINITIONS

- Medial Knee Displacement: Mid-point of the patella passing medially over the great toe during a single leg squat

- Dominant leg: The leg the subject would use to kick a soccer ball for maximal distance.

- Active: Consistently has participated in at least 30 minutes of physical activity three times a week for the past six months.

ASSUMPTIONS

- The use of a digital inclinometer to measure passive range of motion is representative of the antagonists muscle's flexibility.

- The use of a standard goniometer to measure passive range of motion is representative of the antagonists muscle's flexibility.

- The Motion Star Motion Tracking System is a valid measure of joint angles during a dynamic movement.

- The subjects used in this study have similar characteristics to physically active individuals used in other studies, and is representative of the population in which the findings will be generalized to.

- All subjects will truthfully report past and present medical conditions which could exclude them from the study.

- The testing equipment will not inhibit normal body motions.

DELIMINATIONS

- Only subjects who are healthy and active at the University of North Carolina at Chapel Hill will be tested.

- Participants will be categorized based on the criteria established by the researcher.

- People who have had a lower extremity or a low back surgical procedure or have had a lower extremity or low back injury in the past two years which has resulted in three consecutive days of time loss from activity will be excluded.

- All testing will take place in the UNC-CH Neuromuscular Research Laboratory.

LIMITATIONS

- The findings of this study may not be applicable to other populations.

- The individual effort each participant put into correctly completing the SLS and MVIC cannot be assessed.

- The findings of this study are only applicable to the SLS and not other dynamic tasks

- This study did not measure the muscular activation of the other hip, thigh, and lower leg muscles

- EMG electrodes on the skin may not give a true reading of the underlying muscle activity

SIGNIFICANCE OF THIS STUDY

Many certified athletic trainers utilized the single leg squat (SLS) in pre-participation examination to asses an athlete's lower extremity strength and coordination. Through

observation of the single leg squat, many clinicians "diagnose" muscular imbalances/weakness and develop treatment plans to address these. There is little to no research supporting these "diagnoses" through observation; the research which has examined the single leg squat previously has focused on hip abductor strength and knee valgus movement during the single leg squat. This research study will help to serve as a reference to clinicians in identifying problematic lower extremity muscular firing imbalances and lower extremity passive range of motion muscular flexibility while observing the SLS test.

CHAPTER II

LITERATURE REVIEW

INTRODUCTION

Noncontact knee injuries are a common occurrence among physically active populations. These noncontact injuries include injury of the anterior cruciate ligament (ACL) (Noyes, Mooar et al. 1983; Hewett, Lindenfeld et al. 1999; Griffin, Albohm et al. 2006), medial collateral ligament (MCL) (Fetto and Marshall 1978; Grood, Noyes et al. 1981; Griffith, LaPrade et al. 2009; Wijdicks, Griffith et al. 2009), and patellofemoral pain syndrome (PFPS) (Devereaux and Lachmann 1984; Stathopulu and Baildam 2003). These three injuries share one common mechanism, increased knee valgus movement during activity.

Knee valgus collapse during functional activity has been defined as a result of transverse plane knee rotation motions (Hollis, Takai et al. 1991; Quatman and Hewett 2009). Factors associated with knee valgus collapse include: anatomical boney anomalies (Hewett, Stroupe et al. 1996; Boden, Griffin et al. 2000; Griffin, Albohm et al. 2006; Hewett, Myer et al. 2006), lower extremity muscular strength (Hewett, Stroupe et al. 1996; DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006; Hewett, Myer et al. 2006; Lawrence, Kernozek et al. 2008), and lower extremity neuromuscular control (Lloyd and Buchanan 2001; Ford, Myer et al. 2003; Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Hewett, Myer et al. 2005; Myer, Ford et al. 2005). These factors can act

independently or in conjunction with one another to increase knee valgus during both static posture and functional movements.

Injury intervention programs have been developed to aid in counteracting these faulty lower extremity biomechanics. It has been shown that neuromuscular training (Caraffa, Cerulli et al. 1996; Hewett, Lindenfeld et al. 1999; Hewett, Ford et al. 2006; Myer, Ford et al. 2007; DiStefano, Padua et al. 2009), strength training (Hewett, Lindenfeld et al. 1999; Hewett, Ford et al. 2006), and plyometric training are effective in decreasing the risk of ACL injury (Hewett, Stroupe et al. 1996). These prevention strategies focus on teaching proper landing techniques and increasing lower extremity strength and proprioception, in hopes to decrease knee valgus collapse episodes (Caraffa, Cerulli et al. 1996; Hewett, Stroupe et al. 1996; Hewett, Lindenfeld et al. 1999; Hewett, Ford et al. 2006; Myer, Ford et al. 2007; DiStefano, Padua et al. 2009). Recent research suggests that individualized programs based on individual's biomechanical needs may be more effective than group based programs (Myer, Ford et al. 2007; DiStefano, Padua et al. 2009).

Lower extremity functional screening assessments have been suggested to aid in identifying individuals at increased risk of noncontact knee injury (Bonci 1999; DiMattia, Livengood et al. 2005; Newton, Gerber et al. 2006; Hirth 2007). Three-dimensional and two-dimensional video analysis has been utilized in the research setting and shown to be effective in identifying these risk factors (McLean, Walker et al. 2005; Thijs, Van Tiggelen et al. 2007; Ekegren, Miller et al. 2009; Padua, Marshall et al. 2009). These screenings require equipment which is unavailable to many clinicians. Therefore clinically based screenings have been developed. These screenings have gained popularity and are now commonly used

to asses lower extremity tasks (DiMattia, Livengood et al. 2005; Newton, Gerber et al. 2006; Willson, Ireland et al. 2006).

In recent years the single leg squat has gained popularity as one such clinical screening (Zeller, McCrory et al. 2003; Livengood, DiMattia et al. 2004; DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006). However findings of the single leg squat have been based largely on anecdotal evidence and have not been strongly supported by the research. Therefore, additional research is needed to examine the effectiveness of single leg squat as a lower extremity functional clinical assessment. Also, additional research is needed to identify which risk factors can be observed through the single leg squat and what the underlying causes of these injury risk factors are.

ANATOMY OF THE KNEE

In order to have an understanding of knee biomechanics during functional tasks it is important to have a strong understanding of both the dynamic and static anatomical structures which form the knee joint and those structures which act on it. The knee is a diarthrodial synovial hinge joint comprised of the distal femur, proximal tibia, and patella. The medial condyle of the femur is larger than the lateral condyle and projects more distally; this creates an anterior projection of the lateral femoral condyle. These condyles are separated by the femoral notch (Chhabra, Elliott et al. 2001). This arrangement allows for movement in both the sagittal and transverse planes; the knee's primary motions are flexion and extension and its secondary motions are internal and external rotation.

The femur and tibia are separated by the medial and lateral menisci, two crescent shaped fibrocartilaginous structures. These fibrocartilaginous structures help to increase the conformity between the two articulating surfaces and aid in shock absorbency during weight

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bearing activities, especially the more "O-shaped" lateral meniscus (Chhabra, Elliott et al. 2001). The medial meniscus is more "C-shaped" than the lateral meniscus and accommodates the larger medial femoral condyle allowing for greater rotation of the femur on the tibia. Only about 20-30% of the of the peripheral menisci are vascularized (Chhabra, Elliott et al. 2001), limiting the abilities of the menisci to heal after injury. The menisci are connected to the tibial plateau by the coronary ligaments and joined together anteriorly by the transverse ligament (Chhabra, Elliott et al. 2001).

The patella is the largest sesamoid bone in the body and lies in the quadriceps tendon (Chhabra, Elliott et al. 2001). The bone is triangular in shape and sits between the femoral condyles in the femoral groove. The patella serves two major functions in the knee: it acts as a fulcrum for the quadriceps and it is a protective surface for the anterior aspect of the knee joint (Chhabra, Elliott et al. 2001).

The anterior cruciate ligament (ACL) extends from the posteromedial surface of the lateral femoral condyle to the anterior aspect of the intercondyler eminence on the tibia. It is composed of two bundles, an anteromedial bundle and a posterolateral bundle; the anteromedial bundle is taught in flexion while the posterolateral bundle is taught in extension (Chhabra, Elliott et al. 2001). The ACL has been shown to range from 3 to 38 mm in length and 10 to 12 mm in width (Smith, Livesay et al. 1993). Even though the ACL is intraarticualr it is surrounded by its own synovial sheath and derives is blood supply from the middle geniculate artery (Arnoczky 1983). The major function of the ACL is to limit anterior translation of the tibia on the femur; it also aids in limiting rotation of the tibia on the femur (Chhabra, Elliott et al. 2001; Shimokochi and Shultz 2008). As such the ACL is

loaded during anterior tibial translation, transverse plane knee rotations, and valgus collapse (Ferretti, Papandrea et al. 1992; Boden, Griffin et al. 2000; Cochrane, Lloyd et al. 2007).

Similarly the posterior cruciate ligament (PCL) is also intraarticualr and extrasynovial. The PCL extends from the lateral aspect of the medial femoral condyle to a sulcus located posterior and inferior to the tibial plateau. The PCL is also composed of two bundles: the anterolateral bundle and the posteromedial bundle. The anterolateral bundle is taught in flexion while the posteromedial bundle is taught in extension (Chhabra, Elliott et al. 2001). Girgis et al. has reported the average length of the PCL as 38 mm and its average width as 13mm (Girgis, Marshall et al. 1975). The PCL is assisted in preventing posterior translation of the tibia by the ligaments of Humphrey and Wrisberg; which run from the posterior horn of the lateral meniscus to just posterior to the attachment of the PCL on the femoral condyle (Harner, Xerogeanes et al. 1995). The blood supply of the PCL is also provided by the middle geniculate artery (Chhabra, Elliott et al. 2001).

The medial collateral ligament (MCL) acts as the primary static stabilizer of the medial aspect of the knee. Like the ACL and PCL, the MCL is comprised of two portions: the tibial collateral ligament (superficially) and the medial capsular ligament (deep) (Chhabra, Elliott et al. 2001; LaPrade, Engebretsen et al. 2007). Each portion of the MCL has its proximal attachment on the medial femoral epicondyle. The superficial tibial collateral ligament is further divided into anterior and posterior bundles; the anterior bundle has vertically oriented fibers and inserts posterior to the pes anserine tendon, while the posterior bundle has obliquely oriented fibers and inserts inferiorly to the tibial plateau (Chhabra, Elliott et al. 2001). Likewise the medial capsular ligament has two bundles as well, the meniscofemoral and the meniscotibial portions (Chhabra, Elliott et al. 2001;

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LaPrade, Engebretsen et al. 2007). These two bundles attach to the medial meniscus with the coronary ligaments (Chhabra, Elliott et al. 2001) and just distal to the articular cartilage of the tibial plateau (LaPrade, Engebretsen et al. 2007). Unlike the superficial tibial collateral ligament which is an independent structure, the medial capsular ligament is a thickening of the medial joint capsule (LaPrade, Engebretsen et al. 2007). The primary function of the MCL complex is to resist valgus forces applied to the knee and resist rotation of the tibia on the femur (Griffith, LaPrade et al. 2009).

The lateral aspect of the knee is primarily stabilized by the lateral collateral ligament (LCL). The LCL runs from the lateral femoral condyle to the head of the fibula and primarily resists varus forces to the knee, but also resists external rotation of the tibia on the femur. The LCL is assisted in resisting varus force by the iliotibial band (IT band), patellar retinaculum, patellofemoral ligaments, fabellofibular ligament, arcuate ligament, and the lateral joint capsule (Chhabra, Elliott et al. 2001).

The posterolateral corner of the knee is supported by the popliteofibular ligament which connects the fibular head and popliteus tendon. This ligament resists posterior translation, external rotation, and varus forces on the knee. The arcuate, fabellofibular, and oblique popliteal ligaments and the proximal popliteus capsular expansion, part of the posterior capsule, also help resist these motions (Chhabra, Elliott et al. 2001; LaPrade, Morgan et al. 2007).

The primary movers of the knee are the quadriceps and hamstrings. The quadriceps sit on the anterior aspect of the femur and consists of the rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius (Chhabra, Elliott et al. 2001). These four muscles originate from the anterior superior iliac spine (ASIS) and anterior inferior iliac spine (AIIS)

and form a common tendon which attaches to the tibial tuberoscity and act to extend the knee. The hamstrings sit on the posterior aspect of the femur and consists of the biceps femoris (laterally) and the semimembranosus and semitendinosus (medially). These muscles have a common proximal origin off of the ischial tuberoscity and linea aspera of the femur, with individual distal attachments. The short head of the biceps femoris attaches to the lateral tibial condyle while the long head attaches to the fibular head and lateral tibia. The semimembranosus has between five and eight distal points of attachment; these attachments include the oblique popliteal ligament, the posterior capsule, the posterior tibia, the popliteus, and the medial meniscus (Chhabra, Elliott et al. 2001; LaPrade, Engebretsen et al. 2007; LaPrade, Morgan et al. 2007). The semitendinosus joins with the sartorius and gracilis to form the pes anserine tendon which attaches to the anteromedial aspect of the tibia (Chhabra, Elliott et al. 2001).

The popliteus originates on the posterior tibia and inserts on the lateral femoral condyle anterior to the LCL (Chhabra, Elliott et al. 2001). It assists the hamstrings in knee flexion but its primary function is to internally rotate the tibia, in a non-weight bearing position, or to externally rotate the femur in a weight bearing position; in each case it "unlocks" the knee and allows flexion to occur. The popliteus also passively resists hyperexternal rotation of the tibia (Chhabra, Elliott et al. 2001). The gastrocnemius and plantaris muscles act on the knee in a much smaller degree than the other surrounding knee musculature. The gastrocnemius and plantaris muscles primarily act as plantar flexors of the foot but also as weak knee flexors. The gastrocnemius has both medial and lateral heads which along with the semimembranosus and biceps femoris form the medial and lateral borders of the popliteal fossa (Chhabra, Elliott et al. 2001).

The hip adductors also act on the medial knee joint. The adductors have their common distal attachment on the medial femoral condyle, at the adductor tubercle. The adductor magnus has its own separate attachment just posterior and proximal to the adductor tubercle. The adductor magnus tendon fans out and also attaches to the medial gastrocnemius tendon, the posterior oblique ligament, and posteromedial capsule (LaPrade, Engebretsen et al. 2007).

Understanding knee anatomy will allow clinicians to better understand injury that can result from faulty movement patterns. The increased ability to better evaluate knee injuries and movement patterns will allow clinicians to develop individualized prevention and/or rehabilitation programs which better serve their patients.

NONCONTACT KNEE INJURIES

Incidence and Prevalence

Noncontact knee injuries are a common occurrence among physically active populations. In particular physically active people are susceptible to injury of the anterior cruciate ligament (ACL) (Noyes, Mooar et al. 1983; Hewett, Lindenfeld et al. 1999; Griffin, Albohm et al. 2006), medial collateral ligament (MCL) (Fetto and Marshall 1978; Grood, Noyes et al. 1981; Griffith, LaPrade et al. 2009; Wijdicks, Griffith et al. 2009), and to developing patellofemoral pain syndrome (PFPS) (Devereaux and Lachmann 1984; Stathopulu and Baildam 2003). Active persons between the ages of 15 and 25 years are the most commonly afflicted population with ACL injury or PFPS (Devereaux and Lachmann 1984; Griffin, Albohm et al. 2006).

There are an estimated 80,000 to 250,000 ACL injuries annually (Griffin, Albohm et al. 2006). The average cost of surgical repair and subsequent rehabilitation for these injuries

is approximately $17,000 per incident (Hewett, Lindenfeld et al. 1999); this results in a total yearly cost ranging from $1.36-4.25 billion. Both males and females are susceptible to noncontact ACL injuries. However, females sustain 2-8 times more noncontact ACL injuries than males participating in the same sport (Boden, Griffin et al. 2000); this translates to more than 2200 ACL ruptures annually among female collegiate athletes (Hewett, Lindenfeld et al. 1999). The MCL has been reported as one of the most commonly injured ligaments of the knee (Grood, Noyes et al. 1981; LaPrade 1999). The majority of MCL injuries are the result of direct contact to the lateral aspect of the knee with another person or object during activity; however, noncontact MCL injuries are common as well (Hughston, Andrews et al. 1976). Patellofemoral pain syndrome (PFPS) is an encompassing term which refers to all conditions that manifest as pain and point tenderness in or around the patellofemoral joint (Boling, Padua et al. 2009). PFPS affects 1 in 4 people (DeHaven and Lintner 1986; Duffey, Martin et al. 2000), which accounts for up to 25% of all knee related injuries seen in sports medicine clinics (Devereaux and Lachmann 1984). One study of patients suffering from PFPS showed that 36% of all people affected by PFPS must restrict their physical activity because of symptoms associated with the condition (Stathopulu and Baildam 2003).

Injuries to the knee not only affect an individual's physical health, but also the financial cost resulting from these injuries as well as the potential loss of time from competition or the possible loss of scholarship funding can negatively affect a person's mental health (Freedman, Glasgow et al. 1998). These factors make it important to study knee injuries in order to obtain a better understanding of their causes and to identify predisposing risk factors in hopes to form individualized noncontact knee injury intervention programs.

Risk Factors and Mechanisms of Injury

Nearly 70% of ACL injuries result from noncontact mechanisms, the majority of which involve landing from a jump (Boden, Griffin et al. 2000; Hewett, Myer et al. 2006). Inward buckling of the knee has been discouraged during jump-landing and plyometric tasks (Hewett, Stroupe et al. 1996; Hewett, Myer et al. 2005); this is because as the knee buckles inward as a result of a valgus load, the lateral femoral condyle impinges on the ACL. Kennedy et al. (Kennedy, Weinberg et al. 1974) has suggested this impingement of the ACL by the lateral femoral causes a traumatic shearing of the ACL which results in a tearing of the ligament. The connection between dynamic knee valgus during landing and ACL injury has been used as a strategy to help identify those persons who are at risk for future ACL injury (Decker, Torry et al. 2003; Ford, Myer et al. 2003; Hewett, Myer et al. 2005; Kernozek, Torry et al. 2005; Chappell, Creighton et al. 2007; Jacobs, Uhl et al. 2007; Joseph, Tiberio et al. 2008; Lawrence, Kernozek et al. 2008; Boling, Padua et al. 2009; Ekegren, Miller et al. 2009; Kiriyama, Sato et al. 2009; Padua, Marshall et al. 2009).

Females have been shown to have greater total knee valgus motion and greater maximum knee valgus compared to males during vertical jump-landing tasks (Ford, Myer et al. 2003), which increases their susceptibility to noncontact knee injury. Hewett et al. (Hewett, Stroupe et al. 1996) has hypothesized that landing with the knee in either a valgus or varus position places the joint in a less than optimally stable position and in turn makes the knee more susceptible to injury. The increased knee valgus and resulting unstable position has been suggested by Ford et al. (Ford, Myer et al. 2003) as one reason why females have a higher incidence of noncontact ACL injuries compared to males. Ireland et al. (Ireland 1999) termed this position of hip adduction and internal rotation, external rotation of the tibia

relative to the femur, internal rotation of the tibia relative to the foot, and forefoot pronation as "the point of no return." Implying that if a person is placed in this position it is highly likely he/she will damage his/her ACL.

Noncontact MCL injuries most commonly occur during "cutting" maneuvers. Valgus forces applied to the knee during these cutting maneuvers can cause the tibia to externally rotate as the femur remains fixed. This type of valgus injury generally results in tearing of the medial capsular ligament first, and if the force is sufficient enough a subsequent tearing of the tibial collateral ligament is likely as well (Hughston, Andrews et al. 1976).

Many risk factors have been suggested to increase the likelihood of developing PFPS, these include: increased compressive forces on the patellofemoral joint (Boling, Padua et al. 2009), abnormal frontal plane movement of the lower extremity (Willson and Davis 2008), and an increased Q-angle (Reikeras 1992). Many of these risk factors are associated with increased knee valgus during functional tasks. Increased hip internal rotation and knee valgus can greatly increase the lateral compressive forces on the patellofemoral joint (Ireland, Willson et al. 2003; Lee, Morris et al. 2003). An increased Q-angle can result in subsequent increased knee valgus, and any abnormal movement of the femur or tibia in the frontal plane has been suggested to predispose an individual to develop PFPS (Reikeras 1992; Powers 2003).

There are a wide variety noncontact knee injury risk factors and mechanisms of injury. Dynamic knee valgus during functional tasks is a common result of many of these risk factors and often times a key factor in the mechanism of injury. This is why it is

important for clinicians to have a strong understanding of the underlying causes of knee valgus.

KNEE VALGUS

Valgus movement of the knee during functional activities has been disputed through the literature (Hollis, Takai et al. 1991; Quatman and Hewett 2009). Traditionally knee valgus movement has been defined as a pure abduction motion of the tibia relative to the femur (Quatman and Hewett 2009). Quatman et al. (Quatman and Hewett 2009) and Hollis et al. (Hollis, Takai et al. 1991) have further defined dynamic knee valgus, as a valgus collapse episode during functional activity as a result of transverse plane knee rotation motions. Hollis et al. (Hollis, Takai et al. 1991) has described the transverse rotation of the tibia relative to the femur during the application of a valgus load, and found that tibial internal rotation increased as weight bearing knee flexion angles increased. Therefore, functional knee valgus movement is a result of tibial movement in both the frontal and transverse planes (Hollis, Takai et al. 1991; Quatman and Hewett 2009).

When clinically evaluating knee valgus movement during functional tasks, it is often times very difficult, if not impossible, to distinguish what motions at the knee are creating the valgus movement. In order to distinguish between a true valgus episode, where the tibia is abducted from the femur solely in the frontal plane, and an episode where the valgus movement is created by both abduction in the frontal plane and rotation of the tibia in the transverse plane clinicians must use three-dimensional (3-D) motion analysis software to make this distinction. Three-dimensional motion analysis is commonly used to asses knee kinematics during a variety of tasks (Ekegren, Miller et al. 2009; Padua, Marshall et al. 2009). Three-dimensional motion analysis software is used in conjunction with an infrared

optical motion capture system (ie. Vicon) (Myer, Ford et al. 2002; Decker, Torry et al. 2003; Ford, Myer et al. 2003; Hewett, Myer et al. 2005; Kernozek, Torry et al. 2005; Claiborne, Armstrong et al. 2006; Russell, Palmieri et al. 2006; Chappell, Creighton et al. 2007; Heinert, Kernozek et al. 2008; Joseph, Tiberio et al. 2008; Lawrence, Kernozek et al. 2008; Richards, Thewlis et al. 2008; Ekegren, Miller et al. 2009; Kiriyama, Sato et al. 2009) or an electromagnetic motion analysis system (ie. Flock of Birds) (DiMattia, Livengood et al. 2005; Jacobs, Uhl et al. 2007; Boling, Padua et al. 2009; Padua, Marshall et al. 2009); both the motion analysis software and the motion capture system are expensive, time consuming to operate, and unavailable to most clinicians. Motion analysis systems used for the study of knee kinematics have been predominately used to study frontal and sagittal plane movements of the knee during drop-landing tasks (Decker, Torry et al. 2003; Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Lawrence, Kernozek et al. 2008; Kiriyama, Sato et al. 2009), drop-jump tasks (Hewett, Myer et al. 2005; Chappell, Creighton et al. 2007; Joseph, Tiberio et al. 2008; Ekegren, Miller et al. 2009), jump-landing tasks (Jacobs, Uhl et al. 2007; Boling, Padua et al. 2009; Padua, Marshall et al. 2009), the forward lunge (Thijs, Van Tiggelen et al. 2007), single leg squats (Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006; Richards, Thewlis et al. 2008), and running (Heinert, Kernozek et al. 2008).

Two-dimensional (2-D) video analysis of functional activities has been shown to be effective in identifying variations in knee valgus during functional tasks between subjects. The authors agree two-dimensional video analysis has its limitations. The primary limitation of two-dimensional video analysis is that it does not allow for the clinician to properly identify the degree of knee valgus or if there is a rotational component involved (McLean,

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Walker et al. 2005; Ekegren, Miller et al. 2009; Padua, Marshall et al. 2009). Thijs et al. (Thijs, Van Tiggelen et al. 2007) evaluated frontal plane knee movement during a forward lunge and reported that peak knee valgus angle could be properly identified with the use of two-dimensional video analysis. However, Thijs et al. incorporated retroreflective markers to aid in identifying anatomical landmarks on video as well as video analysis software to identify peak knee valgus angles (Thijs, Van Tiggelen et al. 2007).

Valgus movement has been shown to occur as a result of movement of the tibia in both the frontal and transverse planes. The multiple components associated with knee valgus collapse during activity make it a complex motion commonly associated with injury to the ACL, MCL, and other supporting knee structures. Due to the complexity of this motion further study is needed to better understand the motion as a whole and each of its components.

Causes of Knee Valgus

There are a wide variety of factors associated with increased knee valgus. These factors include: anatomical boney anomalies (Hewett, Stroupe et al. 1996; Boden, Griffin et al. 2000; Griffin, Albohm et al. 2006; Hewett, Myer et al. 2006), lower extremity muscular strength (Hewett, Stroupe et al. 1996; DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006; Hewett, Myer et al. 2006; Lawrence, Kernozek et al. 2008), and lower extremity neuromuscular control (Lloyd and Buchanan 2001; Ford, Myer et al. 2003; Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Hewett, Myer et al. 2005; Myer, Ford et al. 2005). These factors can act independently or in conjunction with one another to increase knee valgus during both static posture and functional movements.

Anatomical Factors

The Quadriceps angle, more commonly known as "Q-angle" is a measurement of the vector of pull of the knee extensor musculature and patella tendon. It is measured by drawing a line from the ASIS to the center of the patella, then a second line from the center of the patella to the center of the tibial tuberoscity. This measurement can be done with the subject in either a supine or standing position (Smith, Hunt et al. 2008). Larger than normal Q-angles (males $> 10^o$, females $> 15^0$) have been shown to increase lower extremity malalignment; the most common malalignment is increased knee valgus during a static posture (Boden, Griffin et al. 2000). Females have an increased risk of large Q-angles because of their genetic predisposition of wider hips, compared to males. This has been suggested as one reason as to why females traditionally have greater knee valgus than males during both static and functional tasks (Hewett, Stroupe et al. 1996; Boden, Griffin et al. 2000; Griffin, Albohm et al. 2006; Hewett, Myer et al. 2006).

Increased hip adduction and hip internal rotation have also been associated with increased knee valgus movement during lower extremity closed kinetic chain activities (Hewett, Myer et al. 2005). If the hip is naturally in an adducted position while standing in a neutral position there is commonly an associated increased internal rotation of the hip joint as well. This position can predispose an individual to increased knee valgus during activity since it promotes abnormal rotation of the tibia on the femur associated with the knee valgus collapse described by Hollis et al. and Quatman et al. (Hollis, Takai et al. 1991; Quatman and Hewett 2009).

Lower Extremity Muscular Strength

Lower extremity muscular strength has been suggested by many authors as a contributing factor of knee valgus movement during functional tasks (Hewett, Stroupe et al. 1996; DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006; Hewett, Myer et al. 2006; Lawrence, Kernozek et al. 2008). Traditionally many studies have examined the relationship between hip abductor strength, specifically gluteus medius strength, and knee frontal plane motion; little to no correlation has been found between the two. Claiborne et al. and DiMattia et al. specifically examined the correlation between hip abductor strength and knee valgus movement and hip adduction during a single leg squat. DiMattia's findings agree with past literature and show there is little to no correlation between hip abductor strength and knee valgus movement (DiMattia, Livengood et al. 2005). However, Claiborne et al. (Claiborne, Armstrong et al. 2006) disagree and found a negative correlation between hip abductor strength and knee valgus movement; showing that as hip abductor strength increased, knee valgus movement decreased.

The musculature acting directly on the knee has also been suggested to contribute to knee valgus movement. Hewett et al. has demonstrated in two separate studies that decreased hamstring-to-quadriceps peak torque ratios has led to increased knee valgus movement (Hewett, Stroupe et al. 1996; Hewett, Myer et al. 2006). This decreased hamstring-to-quadriceps ratio is more evident in females than males. Hewett, et al. has attributed this decreased ratio as one factor contributing to greater knee valgus movement in females than in males during functional tasks (Hewett, Stroupe et al. 1996; Hewett, Myer et al. 2006). Even though some correlations have been found between hip and knee muscular strength and predicting knee frontal plane movement, Claiborne et al. found that these

strength factors only account for 22% of the variability in predicting knee frontal plane movement (Claiborne, Armstrong et al. 2006).

Lower Extremity Neuromuscular Control

The relationship between lower extremity muscular control and knee frontal plane movement has been studied extensively. Some studies have suggested hip musculature has an effect on knee frontal plane movement during the SLS (Zeller, McCrory et al. 2003; Hart, Garrison et al. 2007; Lawrence, Kernozek et al. 2008). While others have examined the relationship between the knee's primary movers and knee frontal plane movement (Hewett, Stroupe et al. 1996; Lloyd and Buchanan 2001; Ford, Myer et al. 2003; Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Hewett, Myer et al. 2005; Myer, Ford et al. 2005). Decreased control of either hip or knee musculature (Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Hart, Garrison et al. 2007), imbalances in muscle firing amplitudes of antagonist movers acting on either the hip or knee (Hewett, Stroupe et al. 1996; Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003; Hewett, Myer et al. 2005), or imbalances in firing intensities of the medial and lateral portions of muscle groups acting on either the hip or knee have all been suggested as causes of abnormal knee frontal plane movement (Lloyd and Buchanan 2001; Ford, Myer et al. 2003; Myer, Ford et al. 2005).

Decreased neuromuscular control of either the hip or knee musculature has been hypothesized to be correlated to increased knee valgus movement. Zeller et al. (Zeller, McCrory et al. 2003) showed that females have greater knee valgus movement than males during a SLS task; the author attributes this mainly to what he believes is a decreased control of the gluteus medius muscle found amongst females. The findings of Zeller et al. are supported by similar research that examined gluteus medius activation during a single leg

forward jump task. Hart et al. (Hart, Garrison et al. 2007) found that females had significantly decreased activation of the gluteus medius muscle during a single leg forward jump compared to males. The authors attributed this to decreased control of the gluteus medius muscle and hypothesized this decreased activation was one reason why females exhibited greater hip internal rotation, commonly associated with "the point of no return" (Ireland 1999; Hart, Garrison et al. 2007). Lack of control of the quadriceps muscle has been suggested by DiMattia et al. (DiMattia, Livengood et al. 2005) as one risk factor for increased knee valgus movement during single leg activities.

Imbalances in neuromuscular firing intensities between antagonist movers are predominately seen between the knee's primary movers, the quadriceps and hamstrings. Hewett et al. (Hewett, Stroupe et al. 1996; Hewett, Ford et al. 2006) showed females have decreased hamstring-to-quadriceps peak torque ratios compared to males which the authors believe results in increased knee valgus. Similarly, Hewett et al. (Hewett, Myer et al. 2005) believes if the hamstrings are weak or under recruited during closed kinetic chain knee flexion activities then it could lead to a decrease in quadriceps activation as well; this dual decrease in muscle activation has been hypothesized to lead to decreased dynamic knee stiffness. Decreased dynamic knee stiffness has been associated with unwanted frontal plane knee movement. It has also been suggested that female's increased hamstring flexibility may limit that muscle group's ability to add stability to the knee during functional activities (Boden, Griffin et al. 2000). Electromyography (EMG) studies have shown the rectus femoris to have increased activation in females who went into knee valgus during a SLS (Zeller, McCrory et al. 2003). This supports Hewett et al.'s (Hewett, Stroupe et al. 1996)

theory that males' generally have greater use of the knee flexor musculature during landing tasks which may act as a protective mechanism to prevent knee valgus.

Decreased medial-to-lateral muscular firing ratio during a landing or cutting task can increase knee valgus by compressing the lateral joint and distracting the medial joint surfaces (Lloyd and Buchanan 2001; Ford, Myer et al. 2003; Myer, Ford et al. 2005). Therefore co-contraction of not only the quadriceps and hamstrings collectively but also co-contraction of the medial and lateral components of each muscular group has been suggested to increase frontal plane knee stiffness and aid in preventing knee valgus movement (Hewett, Myer et al. 2005)

There are multiple factors which can contribute to knee valgus movement during activity. Knee valgus movement may be caused by one or all of these factors. It is important to gain a better understanding of these contributory factors and how each factor affects the individual's performance during functional tasks. By better understanding how each factor affects the individual's performance, clinicians may be able to develop injury intervention programs for these valgus contributors.

INTERVENTION PROGRAMS

Injury intervention programs have been developed in hopes of correcting faulty lower extremity biomechanics which predispose individuals to injury and ultimately decreasing the incidence of noncontact knee injuries, in particular noncontact ACL injury. Injury intervention programs have utilized a variety of strategies to correct faulty biomechanics. These strategies include plyometric training (Hewett, Stroupe et al. 1996), strength training (Hewett, Lindenfeld et al. 1999; Hewett, Ford et al. 2006), and neuromuscular training (Caraffa, Cerulli et al. 1996; Hewett, Lindenfeld et al. 1999; Griffin, Albohm et al. 2006;

Myer, Ford et al. 2007; DiStefano, Clark et al. 2009). All three strategies have been shown to be effective in reducing the incidence of noncontact knee injuries; however, neuromuscular training programs have had the greatest success in altering and decreasing faulty lower extremity biomechanics (Caraffa, Cerulli et al. 1996; Hewett, Lindenfeld et al. 1999; Hewett, Ford et al. 2006; Myer, Ford et al. 2007; DiStefano, Clark et al. 2009).

Lower extremity neuromuscular training programs have focused on teaching proper landing techniques, balance, and proprioception. The aim of these programs is to decrease knee valgus collapse episodes commonly associated with noncontact knee injuries during functional tasks (Caraffa, Cerulli et al. 1996; Hewett, Stroupe et al. 1996; Hewett, Lindenfeld et al. 1999; Hewett, Ford et al. 2006; Myer, Ford et al. 2007; DiStefano, Padua et al. 2009). These injury intervention programs have had positive results in decreasing noncontact knee injuries. However, these programs are designed to be implemented to large groups of individuals and therefore assign all participants the same exercises. Recent research suggests individuals in these programs respond differently to them based on their initial biomechanical profile. Those individuals in the program who were labeled as "high risk" had greater improvement than those individuals who were labeled as "low risk" at the beginning of the studies (Myer, Ford et al. 2007; DiStefano, Padua et al. 2009).

Therefore it has been suggested that similar injury intervention programs based on biomechanical preparticipation screenings may be more effective in the reduction of noncontact knee injuries (Myer, Ford et al. 2007; DiStefano, Padua et al. 2009). Ideally each individual in an injury intervention program would undergo a dynamic lower extremity biomechanical assessment prior to participation in the program. By screening each

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individual the intervention programs could be tailored to meet the individual's biomechanical needs and in turn increase the efficacy of the program.

In order for these individualized programs to be successful clinicians must have reliable and validated lower extremity clinical screening assessments. Clinicians must also be able to correctly identify the underlying causes of their findings of these screenings. Therefore it is essential to provide clinicians with reliable screenings and data on how to interpret the findings of these screenings.

LOWER EXTREMITY CLINICAL SCREENING ASSESSMENTS

Lower extremity functional screening assessments have been suggested to aid in identifying athletes at increased risk of noncontact knee injury (Bonci 1999; DiMattia, Livengood et al. 2005; Newton, Gerber et al. 2006; Hirth 2007). Therefore there is a need to develop screening assessments which are quick, easy to complete, and do not require extra equipment to administer the assessment.

Screening tools have gained popularity and are now commonly used to asses parts of lower extremity tasks (DiMattia, Livengood et al. 2005; Newton, Gerber et al. 2006; Willson, Ireland et al. 2006). Hewett et al. (Hewett, Myer et al. 2001) has suggested that when screening for ACL risk factors these clinical assessments should focus on quadriceps dominance, dynamic knee valgus movement during landing, and single leg stance balance. Many of the clinical based assessments evaluate the same or similar tasks as the laboratory based assessments. Clinically based assessments examining knee frontal plane movement have looked at: the overhead squat (Hirth 2007), jump-landing tasks (Padua, Marshall et al. 2009), the forward lunge (Crill, Kolba et al. 2004; Thijs, Van Tiggelen et al. 2007), and the

single leg squat (Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006; Willson and Davis 2008).

The overhead squat test and Landing Error Scoring System (LESS) have been validated as screening tools for identifying lower extremity risk factors for noncontact knee injury. These tests require little space and can be completed in a clinical setting without specialized equipment. They can be effectively carried out by even new clinicians with only minimal instruction (Hirth 2007; Padua, Marshall et al. 2009). Similarly the forward lunge requires little space to complete and can also be completed without specialized equipment. It is becoming a popular screening tool amongst physical therapists and athletic trainers because it is similar to the gait cycle. It too has been shown to be a reliable functional test to asses an individual's ability to move the lower extremity in the frontal and sagittal planes (Crill, Kolba et al. 2004)

The single leg squat (SLS) has gained popularity in recent years as a lower extremity screening assessment. It has traditionally been used to identify individuals with poor hip abductor strength and poor trunk control (Zeller, McCrory et al. 2003). However these assessments have been based almost entirely on anecdotal evidence and not strongly supported by the current literature. Willson, et al. (Willson, Ireland et al. 2006) found a weak correlation between core strength and knee frontal plane movement during a single leg squat. A weak correlation was also found between hip abductor strength and increased knee valgus movement during a single leg squat (Claiborne, Armstrong et al. 2006); but a recent study by DiMattia et al. (DiMattia, Livengood et al. 2005) has shown there is little to no correlation between hip abductor strength and frontal plane knee motion. In a later study Willson and Davis (Willson and Davis 2008) showed the measure of the frontal plane projection angle

(FPPA) during the SLS may be a useful clinical measure to determine an individual's predisposition to develop PFPS (Willson and Davis 2008)

The development of lower extremity clinical screening assessments is vital to aid in decreasing the amount of noncontact knee injuries which affect a large portion of physically active people. These screening assessments must be cost and time effective, require minimal space to complete, mimic lower extremity functional tasks, and be well researched and understood. Thorough research of these assessments will allow clinicians to have a better understanding of what the assessment is telling them and in turn how to better develop rehabilitation and treatment plans.

AREAS OF NEEDED RESEARCH

Additional research continues to be needed in the examination of the single leg squat as a lower extremity functional clinical assessment. The SLS has yet to be validated as a screening tool for identifying noncontact knee injury risk factors (Willson and Davis 2008). Also, if validated, additional research is needed to identify which risk factors can be observed through the SLS. More importantly further research is needed to identify what the underlying causes of these injury risk factors are, because these underlying causes are what should be addressed through individualized rehabilitation programs.

SUMMARY

Noncontact knee injuries are common among physically active people. One of the leading predisposing factors associated with noncontact knee injuries is increased valgus during activity. Laboratory testing has identified a multitude of factors associated with increased knee valgus. Injury intervention programs have been developed to address a number of these factors and have been successful in correcting faulty biomechanics and

decreasing the rate of noncontact injuries. These injury intervention programs traditionally have taken a group approach and assigned all participants the same set of exercises, but recently it has been suggested that individualized programs may be more effective in correcting faulty biomechanics. Therefore it is essential to provide clinicians with low cost, easily administered clinical assessments of lower extremity functionality and guides on how to interpret their findings so clinicians can develop highly effective biomechanical injury intervention programs.

CHAPTER III

METHODS

SUBJECTS

A total of forty individuals (20 males, 20 females) were selected from a larger group of participants who volunteered to participate in this study. All participants were selected from a convenience sample of students, faculty, and staff from the University of North Carolina at Chapel Hill. Each participant was assigned to either the "control" group or "MKD" group based on his/her performance of the single leg squat (SLS) test. Each group contained a total of twenty participants, ten males and ten females; there were no significant differences between groups for height, weight, or age (descriptive statistics available in Table 3.1).

Inclusion Criteria

All study participants were students, faculty, or staff at the University of North Carolina at Chapel Hill, between the ages of 18 and 35 years. All participants self-reported to be in good physical condition and physically active, defined as consistent participation in at least 30 minutes of physical activity, three times a week for the past six months.

Exclusion Criteria

Study participants were excluded from this study if they had any history of a surgical procedure to their lower extremity or low back and/or reported an injury to the lower extremity or low back within the past six months which had resulted in an inability to

participate in physical activity for three consecutive days. Persons with a known neurologic condition resulting in decreased balance and/or proprioception, and knowingly pregnant females were excluded. Participants were also excluded from the study if they went into a knee varus position during their single leg squat group assignment trials.

INSTRUMENTATION

Electromagnetic Tracking System

A Motion Star (Ascension Technologies, Inc, Burlington, VT) electromagnetic motion tracking system was used to track lower extremity kinematics. The device consists of an extended range transmitter that emits an electromagnetic field and standard receivers (dimensions 25.4 X 25.4 X 20.3 mm) that detect the electromagnetic field. The Motion Star system tracked and recorded the position and orientation of the receivers about the *x*, *y*, and *z* axes relative to the transmitter. Electromagnetic tracking systems have been reported to provide accurate (An, Jacobsen et al. 1988; Milne, Chess et al. 1996) and reliable (An, Jacobsen et al. 1988) data for 3-dimensional movement of body segments and joints. These data were used to objectively identify the start position, the point of greatest knee flexion, and the end position as the subject returned to the start position.

Digital Inclinometer

Joint angles for measures of flexibility of the hip external rotators, hip internal rotators, hamstrings, iliotibial band, and iliopsoas were measured using a digital inclinometer (Saunders Group, Inc, Chaska, MN). Intersession and intrarater reliability of the passive range of motion testing procedure of the investigator responsible for taking the measures in this study was calculated with intraclass coefficients (ICC) and standard errors of the

measurement (SEM) for each range of motion measurement ($ICC_{3,1}$ range, .64-.93; SEM range, $.68^0$-7.45^0) (Table 3.2).

Standard Goniometer

Joint angles for measures of flexibility of the hip adductors, plantar flexors, and dorsi flexors were measured using a standard 30.5 cm (12 in) plastic goniometer. Intersession and intrarater reliability of the passive range of motion testing procedure of the investigator responsible for taking the measures in this study was calculated with intraclass coefficients (ICC) and standard error of the measurement (SEM) for each range of motion measurement ($ICC_{3,1}$ range, .82-.90; SEM range, 2.52^0-3.60^0) (Table 3.2).

Electromyography

A surface electromyography (EMG) system (Bagnoli-8; Delsys, Inc, Boston, MA) with an interelectrode distance of 10 mm, amplification factor of 1,000 (20 – 45 Hz), common-mode rejection ratio of 60 Hz (>80 dB), and input impedance > 10^{15}//0.2 Ω//pF was used to record lower extremity muscle activity. Kollmitzer et al. (Kollmitzer, Ebenbichler et al. 1999) showed EMG measures of lower extremity muscle activity to be reliable for short-term and long-term test-retest intervals.

Videography

Two 2-dimensional video cameras (DCR-HC38 MiniDV Handycam Camcorder; Sony Electronics, San Diego, CA) were positioned to capture and record an anterior view and a lateral view of each single leg squat trial so that they could be reviewed at a later date if necessary to observe knee position during the trials.

PROCEDURES

Screening Session

Prior to data collection each participant underwent a screening protocol to determine his/her inclusion into the study and group assignment. The participants read and signed an informed consent form approved by the Institutional Review Board (IRB) of the University of North Carolina at Chapel Hill, and each participant was able to ask questions to clarify any part of the informed consent form prior to signing it. All participants also completed a questionnaire to confirm inclusion/exclusion criteria, the participant's dominant leg (the leg that would be used to kick a soccer ball for maximal distance), and the participant's contact information. Following completion of the questionnaire anthropometric measurements of height (cm) and mass (kg) were taken. All participants wore their own t-shirt and athletic shorts and were barefoot throughout the screening and all data collection.

Participants then completed a warm-up on a stationary cycle ergometer at a self-selected pace for 5-minutes. Following the warm-up the participant underwent a screening protocol to determine group assignment. The screening protocol consisted of each study participant completing five consecutive single leg squat trials while being visually observed by one of the researchers.

Prior to screening each participant performed a single leg squat while having his/her knee flexion angle measured with a standard goniometer. Once the participant reached 60^0 of knee flexion a mechanical block was set so that it just touched the participant's gluteus maximus muscles. Each participant was then instructed to stand on his/her dominant leg, with his/her toes facing forward. The non-weightbearing leg was flexed at the knee to 90^0 and 45^0 at the hip, the hands were placed on the hips, and the head and eyes faced forward.

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The participant then flexed the weightbearing knee to 60^0 as he/she descended for two beats of the metronome until the gluteals touched the mechanical block, then returned to the starting position in two beats. The metronome was set at a frequency of 60 beats/minute.

Participants were placed in the "control" group if in at least three of five trials he/she maintained a neutral knee position (knee remained in line with the hip and ankle joints throughout the single leg squat); participants were placed in the "MKD" group if in at least three of five trials he/she went into medial knee displacement (midpoint of patella moving medially to the great toe during the single leg squat) (Bell, Padua et al. 2008). Selected participants were contacted at a later date to complete data collection.

Lower Extremity Flexibility

On the day of data collection participants again completed the 5-minute warm-up on the cycle ergometer and were rescreened to confirm group assignment. If a participant no longer fit into his/her originally assigned group he/she was excluded from the study, in order to not confound the study results. Lower extremity passive range of motion (ROM) was then measured for each participant in a counterbalanced order to prevent any potential order effect. All ROM measurements were measured with a digital inclinometer or standard goniometer. Intersession and intrarater reliability and precision were established prior to data collection. For each of the following muscle groups the tester passively moved the associated joint through its range of motion from a neutral position to the point of first resistance. The point of first resistance was defined as the point where the tester felt resistance from tension in the muscle and other soft tissue structures, or the participant vocalized discomfort. Three trials were taken for each ROM measurement. The following procedures were utilized for ROM measurements:

- *Hip external rotators*: The participant was positioned in a prone position with his/her knee bent to 90^0, so that the shank and foot were perpendicular to the floor, and the femur was in line with the body; the other leg was flat on the table. One researcher stabilized the participant's pelvis by placing a hand on the sacrum then grasped the shank of the leg to be measured with the opposite hand and passively internally rotated the femur until the point of first resistance. Once this point was reached a second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed perpendicular to the length of the lateral tibia (Starkey and Ryan 2002). (Figure 3.1)

- *Hip internal rotators*: The participant was positioned in a prone position with his/her knee bent to 90^0, so that the shank and foot were perpendicular to the floor, and the femur was in line with the body; the other leg was flat on the table. One researcher stabilized the participant's pelvis by placing a hand on the sacrum, then grasped the shank of the leg to be measured and passively externally rotated the femur until the point of first resistance. Once this point was reached a second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed perpendicular to the length of the medial tibia (Starkey and Ryan 2002). (Figure 3.2)

- *Hip adductors*: The participant was placed in a supine position with his/her legs in full extension, flat on the table. One researcher stabilized the pelvis by placing a hand on the contralateral anterior superior iliac spine (ASIS) of the leg being tested then grasped the medial aspect of the shank on the leg being measured and passively abducted the leg until the point of first resistance. Once this point was reached a second researcher measured the angle with a standard goniometer with the stationary

arm positioned so the distal portion was placed over the contralateral ASIS, the fulcrum over the ipsilateral ASIS, and the movement arm over the long axis of the femur, with the middle of the patella as the distal reference (Starkey and Ryan 2002). (Figure 3.3)

- *Iliopsoas*: The participant was placed in a supine position with his/her hip joint positioned over the edge of the table. He/she flexed the nondominant hip, brought the nondominant knee to the chest and held this position while the low back, sacrum, and pelvis remained horizontal to the table and were stabilized by one researcher; the participant relaxed the contralateral leg and let it drop down toward the table until the point of first resistance. Once this point was reached a second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed along the anterior aspect of the thigh at the midpoint between the ipsilateral ASIS and the patella (Ferber, Kendall et al. 2010). (Figure 3.4)

- *Hamstrings-Leg at 90-90*: The participant was placed in a supine position with the dominant leg flexed to 90^0 of hip flexion and 90^0 of knee flexion (stabilized by one researcher) and the contralateral leg flat on the table. A second researcher grasped the shank of the leg being tested and passively extended the knee until the point of first resistance. Once this point was reached the second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed along the anterior aspect of the tibia (Magee 2006). (Figure 3.5)

- *Iliotibial band*: The participant was placed in a side lying position on his/her nondominant side with the pelvis and shoulders aligned along the vertical plane and his/her dominant knee flexed to 90^0, the contralateral leg was in full extension, flat on

the table with the dominant leg resting on top. One researcher stabilized the pelvis by placing a hand on the ASIS of the leg being tested then grasped the dominant knee on the medial side and moved the thigh into hip flexion, abduction, and extension and then lowered the leg into adduction until the point of first resistance. Once this point was reached a second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed along the lateral aspect of the thigh at the midpoint between the ipsilateral ASIS and patella (Ferber, Kendall et al. 2010). (Figure 3.6)

- *Plantar flexors*: The participant was placed in a supine position with both legs fully extended, the foot being tested was positioned so the ankle extended off the end of the table. One researcher stabilized the shank by grasping the tibia/fibula at mid-shaft and grasped the foot with the opposite hand and passively moved the foot into a dorsi flexed position until the point of first resistance. Once this point was reached he measured the angle with a standard goniometer with the stationary arm aligned with the long axis of the fibula, the fulcrum centered over the lateral malleolus, and the movement arm placed parallel with the long axis of the fifth metatarsal; the procedure was repeated with the knee flexed to a 90^0 angle (Starkey and Ryan 2002). (Figures 3.7 & 3.8)

- *Hip anteversion*: The participant was positioned in a prone position with his/her knee bent to 90^0, so that the shank and foot were perpendicular to the floor, and the femur was in line with the body; the other leg was flat on the table. One researcher stabilized the participant's pelvis by placing a hand on the sacrum. A second researcher palpated the greater trochanter of the dominant leg while passively rotating the hip until the most prominent part of the greater trochanter was in the most lateral

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position. Once this point was the second researcher measured the angle, with respect to the horizontal, with a digital inclinometer placed perpendicular to the length of the medial tibia (Nguyen and Shultz 2007). (Figure 3.9)

- *Posterior talar glide*: The participant was positioned seated so that the legs hung off of the end of the table so the knees were in a flexed position. One researcher established subtalar neutral position with his/her thumbs and maintained the foot in this neutral position. The researcher then applied a posteriorly directed force to the talus until a capsular end-feel was detected. Once this point was reached a second researcher measured the angle, with respect to the vertical, with a digital inclinometer placed along the length of the tibia (Grindstaff, Beazell et al. 2009). (Figure 3.10)

Electromyography

Prior to electrode application, each electrode site was identified and marked with a felt tip marker. Each site was shaved using an electric razor and cleaned with a 70% isopropyl alcohol solution to reduce skin impedance. The following muscles and electrode sites were utilized for the study (Figures 3.11 & 3.12):

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

- *Gluteus medius*: 33% of the distance from the iliac crest to the greater trochanter, starting from the greater trochanter (Rainoldi, Melchiorri et al. 2004)

- *Hip adductors*: medial thigh approximately 2 cm distally from the pubic bone (Cram, Kasman et al. 1998)

- *Medial hamstrings*: 36% of the distance from the ischial tuberoscity to the medial side of the popliteus cavity, starting from the ischial tuberosity (Rainoldi, Melchiorri et al. 2004)

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

- *Vastus medialis oblique*: 20% of the distance from the ASIS to the medial joint space, starting from the joint line (Ericson, Nisell et al. 1985)

- *Vastus lateralis*: 25% of the distance from the ASIS to the lateral joint space, starting from the joint line (Ericson, Nisell et al. 1985)

- *Medial head of gastrocnemius*: 50% of the distance from the medial side of the popliteus cavity to the medial side of the Achilles tendon insertion, starting from the Achilles tendon insertion (Rainoldi, Melchiorri et al. 2004)

Each electrode was placed parallel to the orientation of the muscle fibers; one reference electrode was placed over the anteromedial portion of the proximal tibia. Electrode placement was confirmed with manual muscle testing of each muscle and observation of the muscle activity on an oscilloscope. Once electrode placement was confirmed, the electrodes and leads were secured with clear, plastic surgical tape. Each respective muscle group (hip extensors, hip abductors, hip adductors, hamstrings, quadriceps, and plantar flexors) then underwent testing for maximal voluntary isometric contraction (MVIC); three, 5 second isometric holds, with one minute of rest between trials. The MVIC data were used to normalize all EMG activation amplitude data. This was done by dividing the average MVIC activation, averaged over a one second window during the period of greatest EMG activation,

by the average EMG activation during the descent phase of the single leg squat. All EMG data were collected at 1000 Hz. The following positions were used for MVIC testing:

- *Hip extensors*: The participant was placed in a prone position with the dominant leg flexed at the knee to 90^0 and the contralateral leg lying flat on the table. The researcher stabilized the pelvis by placing a hand on the subject's sacrum. The researcher's other hand was placed over the posterior aspect of the participant's thigh, just proximal to the knee joint line. The participant was instructed to maintain the flexed knee position during testing and to attempt to raise his/her thigh off of the testing table while the researcher applied downward pressure (Kendall FP, McCreary EK et al. 1993). (Figure 3.13)

- *Hip adductors*: The participant was placed in a side lying position on the side of the dominant leg with the dominant leg fully extended and the nondominant leg flexed at the knee and hip, so that the sole of the participant's nondominant foot could be placed on the testing table in front of the dominant leg. The researcher stabilized the pelvis by placing a hand on the subject's nondominant side iliac crest. The researcher's other hand was placed over the medial aspect of the knee, just proximal to the knee joint line. The participant was instructed to maintain a neutral hip rotation position throughout the testing and to attempt raise his/her dominant leg off of the testing table while the researcher applied downward pressure (Kendall FP, McCreary EK et al. 1993). (Figure 3.14)

- *Hip abductors*: The participant was placed in a side lying position on the side of the nondominant leg with both legs fully extended and the dominant leg resting on top of the nondominant leg. The researcher stabilized the pelvis by placing a hand on the

subject's dominant side iliac crest. The researcher's other hand was placed over the lateral aspect of the knee, just proximal to the knee joint line. The participant was instructed to maintain a neutral hip rotation position throughout the testing and to raise his/her dominant leg until it was parallel with the testing table and hold it there while the researcher applied downward pressure (Kendall FP, McCreary EK et al. 1993). (Figure 3.15)

- *Hamstrings*: The participant was placed in a prone position with the dominant leg flexed at the knee to 90^0 and the nondominant leg lying flat on the table. The researcher stabilized the leg to be tested by placing one hand on the distal 1/3 of the posterior aspect of the thigh. The researcher's other hand grasped the posterior aspect of the dominant leg's heel. The participant was instructed to attempt to pull his/her heel in towards his/her gluteal muscles as the researcher resisted the motion with pressure opposing the motion (Anderson, Hall et al. 2005). (Figure 3.16)

- *Quadriceps*: The participant was placed in a seated position with both of his/her legs extending off of the table and flexed at the knee to 90^0 and his/her hands crossed across the chest. The researcher stabilized the thigh to be tested by placing one hand on the distal 1/3 of anterior aspect of the thigh. The researcher's other hand grasped anterior aspect of the participant's shank just proximal to the malleoli.   The participant was instructed to attempt to fully extend his/her knee as the researcher applied downward pressure (Anderson, Hall et al. 2005). (Figure 3.17)

- *Plantar flexors*: The participant was placed in a prone position with both of his/her legs fully extended. The researcher stabilized the shank by grasping the dominant leg at mid-shaft of the tibia/fibula. The other hand was placed on the plantar aspect of

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the foot over the metatarsal heads. The participant was instructed to attempt plantar flex his/her foot while the researcher applied pressure directed towards the participant (Kendall FP, McCreary EK et al. 1993). (Figure 3.18)

Motion Analysis

Each subject was fitted with three electromagnetic sensors placed over the sacrum, lateral aspect of the thigh, and the anteromedial aspect of the proximal tibia. All sensors were placed over the area of least muscle mass and secured with double-sided tape, prewrap, and athletic tape. Six additional boney landmarks were digitized with a fourth electromagnetic sensor placed on the end of a point stylus: medial and lateral femoral condyles, medial and lateral malleoli, and left and right anterior superior iliac spines (ASIS). Joint centers for the ankle and knee were calculated as the midpoints of the respective points based on the position of the medial and lateral malleoli and medial and lateral femoral condyles, respectively; the Bell method was used to approximate the hip joint centers (Bell, Pedersen et al. 1990). Three-dimensional coordinate data were collected at a sampling rate of 100 Hz.

Single Leg Squat Task

Prior to data collection each participant performed a single leg squat while having his/her knee flexion angle measured with a standard goniometer. Once the participant reached 60^0 of knee flexion the mechanical block was set so that it just touched the participant's gluteus maximus muscles. Each participant was then instructed to stand on his/her dominant leg, with the toes facing forward. The non-weightbearing leg was flexed at the knee to 90^0 and 45^0 at the hip, the hands were placed on the hips, and the head and eyes faced forward. The participant then flexed the weightbearing knee to 60^0 as he/she

descended for two beats of the metronome until the gluteals touched the mechanical block, then returned to the starting position in two beats. The metronome was set at a frequency of 60 beats/minute. The participant was also instructed as to what constituted a successful trial; no additional feedback, coaching, or other instruction were provided to the participant concerning technique. A trial was deemed successful if: 1) the participant maintained proper testing position throughout the entire motion; 2) the participant squatted until the gluteals came in contact with the mechanical block; 3) the task was completed at the appropriate rate; 4) the participant did not touch down with the nondominant foot; 5) the legs did not touch together, and the heel maintained contact with the ground and; 6) the task was completed in a fluid motion. The participants were given as many practice trials as needed to perform the task successfully.

Once the participant felt comfortable with the single leg squat task, EMG and motion analysis data were then collected simultaneously for five successful single leg squats. The participant completed as many trials as necessary until five successful trials were recorded. Following each trial the participant was given a one minute break, in which time he/she could relax and place both feet on the ground.

DATA PROCESSING AND REDUCTION

The Motion Monitor Software (Innovative Sports Training, Inc, Chicago, IL) was used to control both the Motion Star and EMG systems. A global coordinate system was established for the Motion Star system where the x-axis corresponded to the participant's antero-posterior axis, the y-axis corresponded to the medio-lateral axis, and the z-axis corresponded to the longitudinal axis; local coordinate systems were established for each sensor once it was placed on the participant and the participant stood in anatomical position.

A right-handed Cartesian coordinate system was used to estimate 3-dimensional (3-D) coordinates of lower extremity bony landmarks. Joint angles were calculated with Euler angles (Euler sequence y, x', z"). Knee flexion/extension was defined as the shank relative to the thigh about the y-axis; knee valgus/varus was defined as the shank relative to the thigh about the x-axis, and tibial internal/external rotation was defined as the shank relative to the thigh about the z-axis. All kinematic data were filtered using a fourth-order low-pass Butterworth filter at 14.5 Hz. EMG data were passively demeaned, bandpass (10-350 Hz) and notch (59.5-60.5 Hz) filtered, and smoothed using a 25 ms root mean squared sliding window function. Kinematic and EMG data were exported to and reduced using a custom Matlab program (Math Works, Natick, MA).

EMG and kinematic data were recorded during the descent phase of the single leg squat; the time from the start of the trial to the point of greatest knee flexion. Three-dimensional joint angles were also recorded at the participant's initial position at the start of the trial (start position: the static position when the participant stood on his/her dominant leg and had the nondominant leg flexed at the knee to 90^0 and 45^0 at the hip and the hands were placed on the hips), the point when the participant reached peak knee flexion, and once the participant returned to his/her initial position (end position). EMG and kinematic data were averaged over the five trials of the single leg squat and range of motion data were averaged from the three trials of each passive range of motion measurement.

STATISTICAL ANALYSIS

To compare normalized EMG mean amplitude over the course of the descent phase of the single leg squat and passive range of motion mean measurements between the control and MKD groups PASW Statistics for Windows software (version 18.0, SPSS Inc, Chicago, IL)

was used to run three separate multivariate analysis of variance (MANOVAs): one MANOVA was run comparing the peak EMG amplitude of the valgus group to the mean peak EMG amplitude of the control group for each of the eight muscles examined in this study; one MANOVA was run comparing the EMG co-activation ratios of gluteus medius to hip adductors and gluteus maximus to hip adductors; one MANOVA was run comparing the comparing the mean passive range of motion measurements of the valgus group to the mean passive range of motion measurements of the control group, for each one of the ten muscle groups examined in this study. Appropriate post hoc one-way between subjects ANOVAs were run for significant MANAOVAs. Statistical significance was set at $\alpha < 0.05$.

CHAPTER IV

RESULTS

HEIGHT, WEIGHT, and AGE

Means, standard deviations, and 95% confidence intervals for all height, weight, and age measures are presented in Table 4.1. Three separate independent-samples t-tests were utilized to compare the means for each measurement between the control and MKD groups. No significant differences were observed between height ($T_{38} = -.250$, $P=.804$), weight ($T_{38} = -.184$, $P=.855$), or age ($T_{38} = .000$, $P=1.00$) between the control and MKD groups.

PASSIVE RANGE OF MOTION (ROM) MEASUREMENTS

Means, standard deviations, and 95% confidence intervals for all passive range of motion measures are presented in Table 4.2. A multivariate analysis of variance (MANOVA) test was utilized to compare the mean ROM dependent variables between the control and MKD groups. A significant multivariate main effect for group was observed for ROM measurements (Wilks' Lambda = .555, F(10, 29) $P=.038$, $\eta^2 = .83$). Follow up one-way ANOVAs revealed significant differences in dorsiflexion range of motion with knee straight ($F_{1,38} = 4.203$, $P=.047$), dorsiflexion range of motion with knee flexed ($F_{1,38} = 4.857$, $P=.034$), and posterior talar glide motion ($F_{1,38} = 7.040$, $P=.012$) between the control and MKD groups. Specifically, individuals in the MKD group displayed significantly lesser dorsiflexion ROM with knee straight (5.5±5.4, control = 8.8±4.7), lesser dorsiflexion ROM with knee flexed (9.5±6.2, control = 14.2±7.3) and greater posterior talar glide

(29.8±4.8, control = 25.7±5.0) in comparison to the control group. No other significant differences were observed.

ELECTROMYOGRAPHY MEASUREMENTS

Means, standard deviations, and 95% confidence intervals for all EMG measures are presented in Table 4.3. A MANOVA was run comparing the mean normalized EMG activation between the control and MKD groups. No significant differences were observed between the normalized EMG activation of the two groups (Wilk's Lambda = .742, F(10, 29) P=.280, η^2 = .49). While not significant, it is worth noting the group differences in hip adductor muscle activation was trending toward significance ($F_{1,38}$ = 3.059, P=.089). Specifically, individuals in the MKD group tended to display greater hip adductor EMG activation (20.1±14.0, control = 13.3±9.8). One MKD subject's EMG data were unable to be used in the MANOVA analysis due to abnormal recording of the hamstring EMG activity.

ELECTROMYOGRAPHY CO-ACTIVATION RATIOS

Means, standard deviations, and 95% confidence intervals for calculated co-activation ratios presented in Table 4.4. Muscle co-activation ratios were calculated for gluteus medius activation and hip adductor muscle activation by dividing the mean gluteus medius activity by the mean hip adductor activity (GMed : Hip Add). In addition, co-activation ratios were calculated for gluteus maximus activity and hip adductor muscle activation by dividing the mean gluteus maximus activity by the mean hip adductor activation (GMax : Hip Add). A ratio resulting in 1.0 would indicate completely balanced muscular activation; ratios resulting in values greater than 1.0 indicate greater activation of the muscle in the numerator (GMed and GMax) compared to the muscle in the denominator (Hip Add). A MANOVA was run comparing the ratio of gluteus medius EMG activation to hip adductor EMG activation

between the control and MKD groups and the ratio of gluteus maximus EMG activation to hip adductor EMG activation between groups. A significant multivariate main effect for group was observed for the co-activation ratios (Wilks' Lambda = .822, F(10,29) P=.027, η^2 = .68). Follow-up one-way ANOVAs revealed significant differences between the ratio of gluteus medius EMG activation to hip adductor EMG activation ($F_{1,38}$ = 5.187, P=.028) and the ratio of gluteus maximus EMG activation to hip adductor EMG activation ($F_{1,38}$ = 8.201, P=.007) between the control and MKD groups. Specifically, individuals in the MKD group displayed significantly lesser gluteus medius EMG activation to hip adductor EMG activation ratio (2.4±1.1, control = 4.5±3.9); indicating greater hip adductor EMG activation compared to gluteus medius EMG activation in the MKD group when compared to the control group. The MKD group also had significantly lesser gluteus maximus EMG activation to hip adductor EMG activation ratio (1.1±.62, control = 2.4±1.8) compared to the control group; greater hip adductor EMG activation compared to gluteus maximus EMG activation in the MKD group when compared to the control group.

POWER AND EFFECT SIZE

Observed power (range, .050-.797) and effect size (range, .01-1.07) for each ROM and EMG measure were calculated and are presented in table 4.5. This information will be discussed to describe the clinical significance of these findings.

CHAPTER V

DISCUSSION

To our knowledge, this is the first study to compare hip, knee, and ankle passive range of motion (ROM) measurements, and hip, thigh, and lower leg muscle activation between individuals presenting with medial knee displacement (MKD group) to those who do not (control group) during a single leg squat task. In summary, our findings revealed that dorsiflexion ROM measurements with the knee straight and flexed were significantly lesser and posterior talar glide was significantly greater in the MKD group compared to the control group. However, no other ROM measurements (hip internal rotation, hip external rotation, hip abduction, hip extension, knee extension, femoral anteversion) were different between groups, which suggests that ROM differences may be isolated to ankle dorsiflexion. Muscle activation amplitude of the gluteus medius, gluteus maximus, hip adductors, vastus medialis, vastus lateralis, medial hamstrings, biceps femoris, and medial gastrocnemius were also not significantly different between the two groups. However, co-activation ratios involving the hip adductors and gluteal musculature were different between groups. Specifically, we calculated the co-activation ratio between gluteus medius activation and hip adductor activation (GMed : Hip Add) and gluteus maximus activation and hip adductor activation (GMax : Hip Add). The GMed : Hip Add and GMax : Hip Add ratios were both significantly lesser in the MKD group compared to the control group. Decreased co-activation ratios in the MKD group indicates these individuals use a more hip adductor dominant activation strategy compared to control subjects. Our combined results suggest the

combination of lesser ankle dorsiflexion ROM with altered hip adductor and gluteal musculature co-activation may contribute to dynamic knee valgus during a single leg squat task; each of these findings will be discussed in greater detail throughout the subsequent paragraphs.

Lesser ankle dorsiflexion ROM in the MKD group supports our hypotheses. Based on these findings we believe decreased dorsiflexion ROM may be a large contributor to MKD during functional tasks. The MKD group was observed to have 37.5% and 33.1% less dorsiflexion ROM with the knee straight (effect size = 0.65) and flexed (effect size = 0.70), respectively. These values represent moderate to large effect sizes and further indicate the importance of these differences. Similar findings have been reported when observing ROM differences between a MKD group and a control group who maintained a neutral knee position, in the frontal plane during a double leg squat (Vesci, Padua et al. 2007; Bell, Padua et al. 2008). Limited dorsiflexion has been proposed to contribute to excessive rearfoot pronation and, in turn, result in compensatory increases in lower extremity internal rotation (DiGiovanni and Langer 2007). Greater lower extremity internal rotation may contribute to dynamic knee valgus (Hollis, Takai et al. 1991). The relationship between decreased dorsiflexion and increased knee valgus during dynamic tasks is further supported by Cortes et al. who reported that subjects had significantly greater knee valgus angle at initial contact and decreased dorsiflexion motion after landing when they performed rearfoot landings compared to self-preferred landing styles (Cortes, Onate et al. 2007). Based on these combined findings it appears that restricted ankle dorsiflexion ROM may be an important factor contributing to MKD across a variety of functional tasks. These findings may have important implications in the design of exercise programs aimed at decreasing MKD.

Decreased ankle dorsiflexion ROM may be due to decreased flexibility of the gastrocnemius/soleus complex and/or restricted posterior talar glide on the tibia (Denegar, Hertel et al. 2002). Posterior glide of the talus within the talocrural joint is a necessary accessory motion to allow for full ankle dorsiflexion ROM. Previous research investigating subjects with chronic ankle instability has revealed decreased ankle dorsiflexion ROM and posterior talar glide in these individuals (Vicenzino, Branjerdporn et al. 2006). We originally hypothesized decreased ankle dorsiflexion ROM in the MKD group may be due to restricted posterior talar glide motion. However, this hypothesis was not supported as MKD subjects demonstrated greater posterior talar glide compared to the control subjects (effect size = 0.84). This finding suggests that decreased ankle dorsiflexion ROM in the MKD group was most likely due to decreased flexibility of the gastrocnemius/soleus complex and not restriced posterior talar glide. The sensitivity of the posterior talar glide test has been questioned based on research demonstrating weak associations between the posterior talar glide test with open and closed kinetic chain measures of ankle dorsiflexion ROM (Cosby and Hertel 2011). Thus, future research is needed to better understand the underlying mechanism (decreased muscle flexibility or restricted posterior talar glide) contributing to decreased dorsiflexion ROM in the MKD subjects.

There were no other observed significant differences in ROM measurements between groups. Greater hip internal rotation ROM (Clark and Lucett 2004; Hirth 2007) and femoral anteversion (Nguyen and Shultz 2007) have been suggested to lead to increased femoral internal rotation during dynamic tasks, which could contribute to MKD. Likewise, tightness of the hip adductors and internal rotators has been theorized to result in greater femoral adduction and internal rotation (Clark and Lucett 2004; Hirth 2007), again contributing to

dynamic knee valgus. Tyler et al. (Tyler, Nicholas et al. 2006) has proposed that the factors contributing to iliotibial band and hip flexor tightness may cause the pelvis to tilt anteriorly and result in internal rotation the femur (Tyler, Nicholas et al. 2006). Greater hamstring flexibility, resulting in lesser activation of the hamstrings, may decrease knee stiffness (Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003) and increase the likelihood of greater MKD during activity. However, our findings indicate the flexibility of these muscle groups did not contribute to the presence of MKD.

Our findings suggest that restricted dorsiflexion ROM appears to be the key factor in predicting MKD during the single leg squat. In our study each participant stood so that his/her foot was fixed on the ground with the toes pointing straight ahead, and we assured the heel remained in contact with the ground throughout each trial. As the subject lowered his/her body to the required 60^0 of knee flexion, dorsiflexion also had to occur at the ankle joint. Limited dorsiflexion ROM would inhibit the tibia from moving forward over the foot and may have caused the subject to compensate for this lack of motion. We speculate that MKD subjects compensated for a lack of sagittal plane ankle motion by increasing frontal and/or transverse plane motion at the foot and up through the kinetic chain. Individuals may have compensated by going into more pronation of the foot, eversion of the talus, and internal rotation of the tibia (DiGiovanni and Langer 2007) thus creating the visual appearance of medial knee displacement. Future research investigating the three-dimensional kinematics of the foot and lower leg is needed to better understand if these compensatory motions actually do occur in those individuals displaying MKD.

We observed no statistically significant differences in EMG activation between groups for all muscles investigated. These findings are in agreement with comparable

research, which has demonstrated few differences in EMG activity during single leg squat tasks; however, these past studies examined EMG differences between sexes and not group assignment based on single leg squat performance. Previous research, looking at differences in muscle activation between sexes during the single leg squat, has shown greater activation of the rectus femoris (Zeller, McCrory et al. 2003) and lesser activation of the gluteus medius (Hart, Garrison et al. 2007) in females, both of which are believed to contribute to increased risk of noncontact ACL injury. Padua et al. reported muscle activation amplitude differences between a group visually displaying MKD and a control group that did not during a double leg squat task. These authors reported 34% greater hip adductor muscle activity in the group displaying MKD (Padua, Bell et al. In review). Similar to this finding, our current study showed the MKD group to have 34% greater EMG activity of the hip adductors during the descent phase of the single leg squat; this finding was not statistically significant, but was trending toward it ($P = .089$). In addition, there were no differences found between groups for the activation of the gluteus medius and gluteus maximus muscles, also similar to Padau et al. (Vesci, Padua et al. 2007; Padua, Bell et al. In review).

Therefore, it is proposed that the relative co-activation between the gluteus medius and gluteus maximus with the hip may contribute to MKD. Increased hip adductor activity which is not offset by associated increases in gluteus medius and gluteus maximus activation may allow for the femur to be pulled into a more adducted and internally rotated position (Padua, Bell et al. In review). This is supported by our calculated co-activation ratios between the gluteus medius and the hip adductors (GMed : Hip Add) and the gluteus maximus and the hip adductors (GMax : Hip Add), both of which revealed significant differences between groups that are accompanied by large effect sizes. The co-activation

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ratio of GMed : Hip Add for the MKD group was 2.4 while the control group's ratio was 4.5 (effect size = 0.84); similarly, the ratios for the MKD and control groups for the GMax : Hip Add co-activation ratio were 1.1 and 2.4, respectively (effect size = 1.07). The co-activation ratios were calculated by dividing the gluteal muscle activation (GMed or GMax) by Hip Add activation. Larger co-activation ratios indicate that the GMed or GMax were more active relative to the Hip Add. Conversely, smaller co-activation ratios indicate greater reliance on the Hip Add muscles. Our findings indicate the MKD group places greater reliance on their Hip Add musculature compared to the control group. It is generally thought that MKD may be caused by decreased gluteal muscle strength (Claiborne, Armstrong et al. 2006) or activation (Hart, Garrison et al. 2007). Our findings may help refine this current theory and suggest that MKD may be caused by greater reliance on the Hip Add muscles rather than weakness or decreased activation of the gluteal musculature. We believe that increased hip adductor relative to GMed and GMax activation played a role in facilitating visual MKD during the single leg squat.

There were no differences observed in hip or knee ROM measurements, but there were differences in ankle dorsiflexion ROM. We therefore propose that the imbalance we observed in hip adductor to gluteal activation stems from a neuromuscular compensation as a result of the decreased ankle dorsiflexion ROM. Decreased ankle dorsiflexion ROM creates an abnormal axis of rotation of the tibia on the talus, resulting from altered arthokinematics which limit roll and glide between the joint surfaces. This abnormal rotation applies abnormal stresses on the tissues which have been suggested to produce altered proprioceptive input, which in turn causes the motor control system to adapt (Denegar and Miller 2002). We propose one such altered motor control response is using more frontal and transverse

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plane motion when sagittal plane motion is restricted, resulting in the leg being pulled actively inward. One possible neuromuscular mechanism utilized to achieve this is altering the adductor and gluteal co-activation ratios.

Our findings of lesser dorsiflexion ROM measurements and smaller co-activation ratios as potential factors contributing to MKD may have important implications on injury prevention and rehabilitation programs aimed at decreasing MKD. Individuals displaying lesser dorsiflexion range of motion may benefit from increasing gastrocnemius and soleus flexibility. Also, the clinician could utilize an inhibition technique, such as self-myofascial release (ie. foam rolling), to decrease muscle spindle activity and allow the muscle to relax and be further stretched (Hirth 2007). The clinician could also apply a similar treatment to improve the GMed : Hip Add and GMax : Hip Add ratios by inhibiting the hip adductors. In addition the clinician could utilize rehabilitation exercises focused on increasing gluteus medius and gluteus maximus muscle activation and improving neuromuscular control.

LIMITATIONS

The following limitations should be considered when interpreting the findings of our study. First, our findings are limited to a single leg squatting task as we did not incorporate other functional tasks in our investigation of range of motion and muscle activation. Future research should look at whether findings carryover to when individuals perform more challenging dynamic tasks (ie. jump-landing or cutting maneuvers). Also, our findings are limited to healthy, physically active individuals who display visual MKD during a single leg squat and those who did not; therefore, they may not be applicable to an injured population. We cannot speculate if these individuals would display knee MKD during other tasks. In addition, other lower extremity muscles not investigated in this study could be involved in

dynamic control during a single leg squat. We are also only able to speculate if MKD during a single leg squat is indicative of increased injury risk during physical activity.

Inherent limitations exist with the use of surface EMG. The assumption was made based on previous literature that EMG signal amplitudes represent levels of muscle activity. Crosstalk may occur with the placement of the EMG surface electrodes on the skin and may not give a true reading of the underlying muscle activity. However, we minimized the potential for error by using standard methods of applying the electrodes, sufficiently securing the electrodes to prevent movement, and checking the output of the electrodes prior to data collection to ensure proper placement. Finally, interpretation of our results was based on EMG signals normalized to maximal isometric voluntary activity (MVIC). Another assumption was made that all participants gave their maximal effort during the MVIC measurements and during the single leg squat; this would affect the normalized percentages used during the statistical analyses.

Another potential limitation of our study is that our measure of posterior talar glide was dependent on the investigator's ability to subjectively determine subtalar neutral position and the end feel/restriction in motion as the knee was moved into flexion. However, the investigator responsible for all ROM measurements established himself to have good reliability and precision with this measure (ICC = .93, SEM = 1.2^{0}); therefore, we do not believe this limitation was a major issue with data collection. Future research should look at a more sensitive measure of quantifying restricted posterior talar glide as a possible factor limiting dorsiflexion ROM. Use of an ankle arthrometer to quantify posterior talar displacement and stiffness has been described in previous literature and may be a good tool for future research investigating factors associated with MKD.

CONCLUSION

In conclusion, our findings indicate that dorsiflexion ROM measurements were lesser in subjects displaying MKD compared to those who did not. We believe this limited dorsiflexion may result in compensatory movements in the ankle and lower leg, resulting in foot pronation and tibial internal rotation. Greater levels of hip adductor activity without an associated increase in gluteus medius and/or gluteus maximus activity may increase femoral adduction and internal rotation (Padua, Bell et al. In review); potentially increasing MKD during dynamic tasks. MKD is suggested to be a biomechanical factor associated with anterior cruciate ligament injury, medial collateral ligament injury, and patellofemoral pain syndrome. Rehabilitation and injury prevention programs that increase dorsiflexion, decrease hip adductor activity, and increase hip abductor and external rotator activity may potentially decrease the incidence of these injuries.

Table 3.1 Height (cm), Weight (kg), and Age (yr) Presented as Mean ± SD with 95% Confidence Intervals

Variable	Control		Valgus	
	Mean ± SD	95% CI	Mean ± SD	95% CI
Height, Weight, Age				
Height	173.1 ± 10.1	(168.3, 177.8)	173.8 ± 8.8	(169.7, 177.9)
Weight	71.0 ± 14.6	(64.1, 77.8)	71.8 ± 14.7	(64.9, 78.7)
Age	20.2 ± 1.5	(19.5, 20.9)	20.2 ± 1.8	(19.3, 21.1)

Table 3.2 Intraclass Correlation Coefficients and Standard Error of the Measurement for Passive Range of Motion Measurements

Variable	ICC	SEM
Passive Range of Motion		
Hip External Rotators	.894	4.5
Hip Internal Rotators	.644	7.5
Femoral Anteversion	.731	.68
Iliotibial Band	.752	3.4
Hip Adductors	.906	5.5
Iliopsoas	.898	4.0
Hamstrings (90-90)	.853	5.9
Dorsiflexion (straight)	.821	3.6
Dorsiflexion (flexed)	.904	2.5
Talar Glide	.931	1.2

Table 4.1 Height (cm), Weight (kg), and Age (yr) Presented as Mean ± SD with 95% Confidence Intervals

Variable	Control		Valgus	
	Mean ± SD	95% CI	Mean ± SD	95% CI
Height, Weight, Age				
Height	173.1 ± 10.1	(168.3, 177.8)	173.8 ± 8.8	(169.7, 177.9)
Weight	71.0 ± 14.6	(64.1, 77.8)	71.8 ± 14.7	(64.9, 78.7)
Age	20.2 ± 1.5	(19.5, 20.9)	20.2 ± 1.8	(19.3, 21.1)

Table 4.2 Passive Range of Motion Measurements (degrees) Presented as Mean ± SD with 95% Confidence Intervals

Variable	Control		Valgus	
	Mean ± SD	95% CI	Mean ± SD	95% CI
Passive Range of Motion				
Hip External Rotators	54.7 ± 9.9	(50.0, 59.3)	59.6 ± 8.1	(55.8, 63.4)
Hip Internal Rotators	69.9 ± 12.0	(64.3, 75.5)	72.4 ± 7.3	(69.0, 75.8)
Femoral Anteversion	3.5 ± 3.1	(2.1, 4.9)	3.2 ± 2.8	(1.9, 4.5)
Iliotibial Band	9.9 ± 4.8	(7.6, 12.1)	9.3 ± 4.6	(7.1, 11.4)
Hip Adductors	57.4 ± 9.1	(53.2, 61.7)	53.0 ± 9.9	(48.4, 57.7)
Iliopsoas	21.3 ± 9.9	(16.7, 25.9)	21.4 ± 5.2	(19.0, 23.9)
Hamstrings (90-90)	73.6 ± 12.3	(67.4, 79.8)	69.6 ± 10.8	(64.7, 74.8)
Dorsiflexion (straight)*	8.8 ± 4.7	(6.6, 11.0)	5.5 ± 5.4	(3.0, 8.1)
Dorsiflexion (flexed)*	14.2 ± 7.3	(10.8, 17.6)	9.5 ± 6.2	(10.8, 17.6)
Talar Glide*	25.7 ± .50	(23.4, 28.1)	29.8 ± 4.8	(27.6, 32.1)

*Statistically significant ($P < 0.05$)

Table 4.3 Normalized EMG Measurements Presented as % MVIC ± SD with 95% Confidence Intervals

Variable	Control		Valgus	
	Mean ± SD	95% CI	Mean ± SD	95% CI
Muscle				
Gluteus Maximus	19.5 ± 8.7	(15.5, 23.6)	17.1 ± 9.1	(12.7, 21.5)
Gluteus Medius	37.1 ± 17.3	(29.0, 45.2)	32.9 ± 17.2	(24.6, 41.1)
Hip Adductors	13.3 ± 9.8	(8.7, 17.9)	20.1 ± 14.0	(13.3, 26.8)
Medial Hamstrings	24.7 ± 16.0	(17.2, 32.2)	29.6 ± 16.3	(21.8, 37.4)
Biceps Femoris	30.1 ± 17.8	(21.8, 38.4)	44.4 ± 39.4	(25.5, 63.4)
Vastus Medialis	66.6 ± 26.3	(54.3, 78.9)	70.4 ± 24.8	(58.4, 82.3)
Vastus Lateralis	69.0 ± 23.7	(57.9, 80.1)	75.2 ± 29.1	(61.2, 89.2)
Medial Gastrocnemius	20.4 ± 16.8	(12.5, 28.2)	14.9 ± 18.8	(5.9, 24.0)

Table 4.4 Muscular Co-Activation Ratios Presented as Mean ± SD with 95% Confidence Intervals

Variable	Control		Valgus	
	Mean ± SD	95% CI	Mean ± SD	95% CI
Co-Activation Ratio				
GMed : Hip Adductors*	4.5 ± 3.9	(2.7, 6.3)	2.4 ±1.1	(1.5, 3.2)
GMax : Hip Adductors*	2.4 ± 1.8	(1.5, 3.2)	1.1 ± 0.62	(0.79, 1.4)

*Statistically significant ($P < 0.05$)

Table 4.5 Power and Effect Size of PROM and EMG Measurements

Variable	Effect Size	Observed Power
Passive Range of Motion		
Hip External Rotators	.54	.385
Hip Internal Rotators	.26	.121
Femoral Anteversion	.10	.063
Iliotibial Band	.13	.069
Hip Adductors	.46	.299
Iliopsoas	.01	.050
Hamstrings (90-90)	.35	.165
Dorsiflexion (straight)	.65	.515
Dorsiflexion (flexed)	.70	.575
Talar Glide	.84	.734
Muscle		
Gluteus Maximus	.27	.133
Gluteus Medius	.24	.116
Hip Adductors	.57	.399
Medial Hamstrings	.30	.154
Biceps Femoris	.50	.302
Vastus Medialis	.15	.073
Vastus Lateralis	.23	.111
Medial Gastrocnemius	.31	.155
Co-Activation Ratio		
GMed : Hip Adductors	.84	.603
GMax : Hip Adductors	1.07	.797

Figure 1.1 Control Group Subject   Figure 1.2 Valgus Group Subject

Figure 3.1 Hip External Rotators Passive Range of Motion Measurement

Figure 3.2 Hip Internal Rotators Passive Range of Motion Measurement

Figure 3.3 Femoral Adductors Passive Range of Motion Measurement

Figure 3.4 Iliopsoas Passive Range of Motion Measurement

Figure 3.5 Hamstring Passive Range of Motion Measurement (Leg at 90-90)

Figure 3.6 Iliotibial Band Passive Range of Motion Measurement

Figure 3.7 Plantar Flexors Passive Range of Motion Measurement (knee extended)

Figure 3.8 Plantar Flexors Passive Range of Motion Measurement (knee flexed)

Figure 3.9 Hip Anteversion Meaurement

Figure 3.10 Posterior Talar Glide Measurement
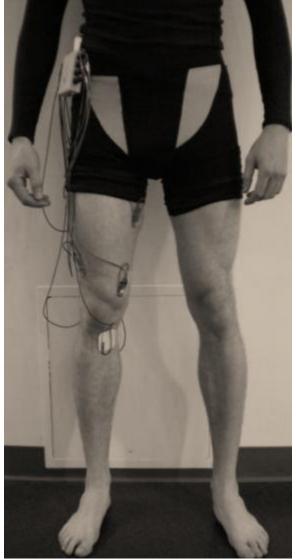
Figure 3.11 Anterior View of EMG
Electrode Sites

Figure 3.12 Posterior View of EMG
Electrode Sites

Figure 3.13 Hip Extensors MVIC Testing Position

Figure 3.14 Hip Adductors MVIC Testing Position

Figure 3.15 Hip Abductors MVIC Testing Position

Figure 3.16 Hamstrings MVIC Testing Position

Figure 3.17 Quadriceps MVIC Testing Position

Figure 3.18 Plantar Flexors MVIC Testing Position

APPENDIX ONE – MANUSCRIPT DRAFT

Title: Lower extremity muscle activation and muscular flexibility and their effect on single leg squat performance

Article Type: Research Paper

Keywords: Single leg squat, knee valgus, dorsi flexion, hip adductor, gluteus medius, gluteus maximus

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Lower extremity muscle activation and muscular flexibility and their effect on single leg squat performance

Context: Knee valgus is a potential risk factor for lower extremity (LE) injuries. LE movement screenings and flexibility measurements may be utilized to identify neuromuscular patterns, which contribute to knee valgus. There are few studies that have investigated how flexibility and muscular activation differ between individuals who display knee valgus during a single leg squat (SLS) and those who do not. We hypothesized that flexibility and muscular activation would differ between the groups.

Objective: To compare LE muscle activation and flexibility between subjects who display visual knee valgus and those who do not during a SLS

Design: Cross-sectional study.

Setting: Neuromuscular research laboratory.

Patients or Other Participants: 40 physically active adults (20 control, 20 valgus)

Main Outcome Measure(s): Subjects completed ten LE flexibility measurements and a five SLS tasks while EMG data were collected from the GMax, GMed, Hip Add, medial hamstrings, biceps femoris, VMO, VL, and medial gastrocnemius. Three MANOVAs were run comparing flexibility measurements, EMG data, and muscle co-activation ratios.

Results: The valgus group had significantly less dorsiflexion ($P = .047$ and $P = .034$), greater talar glide motion ($P = .012$), and smaller gluteus medius to hip adductor co-activation ratio ($P = .028$) and gluteus maximus to hip adductor co-activation ratio ($P = .007$) compared to the control group. There were no differences in individual muscle activation between the two groups

Conclusions: Knee valgus during a SLS appears to be influenced by decreased dorsiflexion and decreased co-activation of the Hip ADD and GMed and GMax muscles. Therefore, rehabilitation and injury prevention programs that increase dorsiflexion, decrease hip adductor activity, and increase hip abductor and external rotator activity may potentially decrease the incidence of these injuries.

Key Words: Single leg squat, knee valgus, dorsi flexion, hip adductor, gluteus medius, gluteus maximus

INTRODUCTION

Noncontact knee injuries are common among physically active populations. In particular, physically active people are susceptible to injury of the anterior cruciate ligament (ACL) (Noyes, Mooar et al. 1983; Hewett, Lindenfeld et al. 1999; Griffin, Albohm et al. 2006), and medial collateral ligament (MCL) (Fetto and Marshall 1978; Griffith, LaPrade et al. 2009; Wijdicks, Griffith et al. 2009), and to developing patellofemoral pain syndrome (PFPS) (Devereaux and Lachmann 1984; Stathopulu and Baildam 2003). Therefore it is important to identify those individuals at increased risk of noncontact knee injury through the use of effective clinical screening instruments that provide insight to the direction of injury prevention and rehabilitation programs.

Malalignment of lower extremity segments during functional activities has been associated with an increase risk of sustaining a noncontact knee injury (Hewett, Stroupe et al. 1996; Boden, Griffin et al. 2000; Griffin, Agel et al. 2000; Griffin, Albohm et al. 2006; Hewett, Ford et al. 2006). One of the most common predisposing factors for noncontact knee injuries is greater knee valgus motion during functional tasks. (Hughston, Andrews et al. 1976; Hewett, Stroupe et al. 1996; Ford, Myer et al. 2003; Hewett, Myer et al. 2005).

In an effort to combat the high incidence of noncontact knee injuries, especially injuries to the ACL, a number of lower extremity screening tools have been developed for identifying injury risk factors associated with these injuries (Bonci 1999; Hewett, Myer et al. 2001; Crill, Kolba et al. 2004; DiMattia, Livengood et al. 2005; McLean, Walker et al. 2005; Newton, Gerber et al. 2006; Hirth 2007; Willson and Davis 2008; Padua, Marshall et al. 2009). The single leg squat has been utilized to identify persons with faulty lower extremity biomechanics (Claiborne, Armstrong et al. 2006), primarily through the observation of

medial knee displacement (MKD). Traditionally, greater MKD observed during the single leg squat has been attributed to poor hip strength or muscle imbalances and poor neuromuscular control of key hip and trunk musculature (Zeller, McCrory et al. 2003; DiMattia, Livengood et al. 2005; Willson, Ireland et al. 2006).

Muscular imbalance has been implicated as a contributing factor to knee valgus. Many studies have examined the influence of musculature surrounding the hip on knee valgus movement during a single leg squat (SLS) (Zeller, McCrory et al. 2003; Lawrence, Kernozek et al. 2008). Strength of the hip abductor group, has received considerable attention in the literature regarding its potential influence on frontal plane knee motion, specifically the gluteus medius; however, recent studies have shown little to no correlation between the two (DiMattia, Livengood et al. 2005; Claiborne, Armstrong et al. 2006). Therefore, other factors such as lower extremity muscular activation and lower extremity flexibility may play an important role in dynamic valgus.

The activation of the musculature surrounding the hip and knee appears to play a major role in the kinematics of the knee joint during functional activities. A relationship has been established between lesser gluteus medius activation and greater knee valgus movement during a single leg squat (Zeller, McCrory et al. 2003). The musculature acting on the knee has been shown to effect knee frontal plane motion. Palmieri-Smith et al. reported greater peak knee valgus angles during a single leg forward-jump task in persons with greater activity in the vastus lateralis and biceps femoris, and lesser peak knee valgus angles in those persons with heightened vastus medialis activity (Palmieri-Smith, Wojtys et al. 2008). The medial gastrocnemius has been suggested to act as a dynamic stabilizer of the knee and helps offset knee valgus moment (Lloyd and Buchanan 2001). Therefore decreased activation of

the medial gastrocnemius during functional tasks may result in decreased frontal plane stability and result in greater knee valgus. No previous research has examined the relationships between the activation of the gluteals, hip adductors, hamstrings, quadriceps, and medial head of the gastrocnemius and their influence on lower extremity kinematics during a single leg squat task.

Lower extremity muscular flexibility has also been associated with greater knee instability and dysfunctional kinematics. Greater flexibility of the hamstrings resulting in lesser activation has been suggested to decrease dynamic knee stiffness (Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003) and increase the likelihood of greater knee valgus. It has been suggested that the factors that contribute to tight iliotibial band and hip flexors may cause the pelvis to tilt anteriorly, potentially resulting in femoral internal rotation (Tyler, Nicholas et al. 2006). Tightness of the hip adductor and internal rotator muscles and increased flexibility of the hip external rotators may allow for greater hip internal rotation and further contribute to greater knee valgus during dynamic tasks. (Clark and Lucett 2004; Hirth 2007) Similarly, tightness of the lower-leg musculature, especially the lateral gastrocnemius and peroneals, may contribute to tibial abduction and external rotation thus increasing greater knee valgus (Clark and Lucett 2004; Hirth 2007). It has been suggested decreased posterior talar mobility would result in decreased dorsi flexion and therefore should be assessed when assessing plantar flexor flexibility (Denegar, Hertel et al. 2002). However, no previous studies have examined the relationships between the flexibility of the hip external rotators, hip internal rotators, hip adductors, hamstrings, iliopsoas, plantar flexors, and talar glide mobility and their effect on lower extremity kinematics during a single leg squat.

Therefore the purpose of this study is to compare the lower extremity muscle activation and flexibility between subjects who display dynamic knee valgus and those who do not during a single leg squat. Determining these differences will aid clinicians by identifying the neuromuscular patterns that are associated with different movement patterns that can be discriminated using cost effective clinical assessments and thereby provide a guide for future rehabilitation interventions to correct these faulty mechanics. Our first hypothesis was that the MKD group will have one or more lower extremity (LE) muscular groups with a significantly higher mean amplitude compared to the control group during a single leg squat. Our second hypothesis was that the MKD will have would have one or more decreased LE passive range of motion patterns compared to the control group.

METHODS

SUBJECTS

Forty individuals (20 males, 20 females) were selected from a larger group of participants who volunteered to participate in this study. Each participant was assigned to either the "control" group or "medial knee displacement" ("MKD") group based on his/her performance of the single leg squat (SLS) test. Each group 20 subjects (10 males and 10 females). Descriptive statistics are available in Table 1.

All participants were self-reported to be in good physical condition and physically active, defined as consistent participation in at least 30 minutes of physical activity, three times a week for the past six months. Subjects were excluded if they had any history of a surgical procedure to their lower extremity or low back and/or reported an injury to the lower extremity or low back within the past six months which had resulted in an inability to participate in physical activity for three consecutive days. Persons with a known neurologic

condition resulting in decreased balance and/or proprioception, and knowingly pregnant females were excluded. Participants were also excluded from the study if they went into a knee varus position during their single leg squat group assignment trials. The participants read and signed an informed consent form approved by the University's Institutional Review Board (IRB).

Instrumentation

A Motion Star (Ascension Technologies, Inc, Burlington, VT) electromagnetic motion tracking system was used to track lower extremity kinematics. These data were used to objectively identify the start position, the point of greatest knee flexion, and the end position as the subject returns to the start position. Joint angles for measures of flexibility of the hip external rotators, hip internal rotators, hamstrings, iliotibial band, and iliopsoas were measured using a digital inclinometer (Saunders Group, Inc, Chaska, MN). Joint angles for measures of flexibility of the hip adductors, plantar flexors, and dorsi flexors were measured using a standard 30.5 cm (12 in) plastic goniometer. Intersession and intrarater reliability of the passive range of motion testing procedure of the investigator responsible for taking the measures in this study was calculated with intraclass coefficients (ICC) and standard error of the measurement (SEM) for each range of motion measurement ($ICC_{3,1}$ range, .64-.93; SEM range, $.68^0$-7.45^0) (Table 3.2). A surface electromyography (EMG) system (Bagnoli-8; Delsys, Inc, Boston, MA) was used to record lower extremity muscle activity. Two 2-dimensional video cameras (DCR-HC38 MiniDV Handycam Camcorder; Sony Electronics, San Diego, CA) were positioned to capture and record an anterior view and a lateral view of each single leg squat trial so that they could be reviewed at a later date if necessary.

Screening Session

Prior to data collection each participant underwent a screening protocol to determine group assignment. Participant's height (cm), mass (kg), and leg dominance (the leg that would be used to kick a soccer ball for maximal distance) were recorded.

Participants completed a warm-up on a stationary cycle ergometer at a self-selected pace for 5-minutes. Following the warm-up the participant underwent a screening protocol to determine group assignment. The screening protocol consisted of each study participant completing five consecutive single leg squat (SLS) trials, to a preset depth of 60^0 of knee flexion (mechanical block set to touch bottom of gluteus maximus during the SLS) while being visually observed by one of the researchers. Each participant was instructed to stand on his/her dominant leg, with his/her toes facing forward. The non-weightbearing leg was flexed at the knee to 90^0 and 45^0 at the hip, the hands were placed on the hips, and the head and eyes faced forward. The participant descended for two beats of the metronome until the gluteals touched the mechanical block, then returned to the starting position in two beats. The metronome was set at a frequency of 60 beats/minute.

Participants were placed in the "control" group if in at least three of five trials his/her knee remained in line with the hip and ankle joints throughout the SLS; participants were placed in the "MKD" group if in at least three of five trials the midpoint of his/her patella moved medially to the great toe during the SLS (Bell, Padua et al. 2008). Selected study participants then went through the following testing procedures at a later date (if a participant did not fit into his/her originally assignment group he/she was excluded from the study).

Lower Extremity Flexibility

Lower extremity passive range of motion (PROM) was measured for each participant in a counterbalanced order. All PROM measurements were measured with a digital inclinometer or standard goniometer. For each of the following muscle groups the tester passively moved the associated joint through its range of motion from a neutral position to the point of first resistance, or the participant vocalized discomfort. Three trials were taken for each ROM measurement. The following procedures were utilized for PROM measurements:

- *Hip external rotators*: The participant was positioned in a prone position with his/her knee bent to 90^0, the femur was then passively internally rotated. The angle was then measured with a digital inclinometer placed perpendicular to the length of the lateral tibia (Starkey and Ryan 2002). (Figure 3.1)

- *Hip internal rotators*: The participant was positioned in a prone position with his/her knee bent to 90^0, the femur was then passively externally rotated. The angle was then measured with a digital inclinometer placed perpendicular to the length of the lateral tibia (Starkey and Ryan 2002). (Figure 3.2)

- *Hip adductors*: The participant was placed in a supine position with his/her legs in full extension, the leg being tested was abducted. The angle was then measured, the leg until the point of first resistance. The angle with a standard goniometer (Starkey and Ryan 2002). (Figure 3.3)

- *Iliopsoas*: The participant was placed in a supine position and completed a Thomas Test for iliopsoas tightness. The angle was measured with a digital inclinometer placed along the anterior aspect of the thigh (Ferber, Kendall et al. 2010). (Figure 3.4)

- *Hamstrings-Leg at 90-90*: The participant was placed in a supine position with his/her dominant leg flexed to 90^0 of hip flexion and 90^0 of knee flexion and the contralateral leg flat on the table; the knee was then passively extended. The angle was measured with a digital inclinometer placed along the anterior aspect of the tibia (Magee 2006). (Figure 3.5)

- *Iliotibial band*: The participant was placed on his/her nondominant side and an Ober's Test for iliotibial band tightness was completed. The angle was measured with a digital inclinometer placed along the lateral aspect of the thigh (Ferber, Kendall et al. 2010). (Figure 3.6)

- *Plantar flexors*: The participant was placed in a supine position with both legs fully extended, the foot being tested was positioned so the ankle extended off of the end of the table; the foot was passively moved dorsi flexion. The angle was measured with a standard goniometer. The procedure was repeated with the knee flexed to a 90^0 angle (Starkey and Ryan 2002). (Figures 3.7 & 3.8)

- *Hip anteversion*: The participant was positioned in a prone position and Clarke's Test for hip anteversion was completed. The angle was measured with a digital inclinometer placed perpendicular to the length of the medial tibia (Nguyen and Shultz 2007). (Figure 3.9)

- *Posterior talar glide*: The participant was positioned in a seated position so that his/her legs hung off of the end of the table so the knees were in a flexed position. Subtalar neutral position was determined and the researcher applied a posteriorly directed force to the talus until a capsular end-feel was detected. The angle was

measured with a digital inclinometer placed along the length of the tibia (Grindstaff, Beazell et al. 2009). (Figure 3.10)

Electromyography

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

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

- *Gluteus medius*: 33% of the distance from the iliac crest to the greater trochanter, starting from the greater trochanter (Rainoldi, Melchiorri et al. 2004)

- *Hip adductors*: medial thigh approximately 2 cm distally from the pubic bone (Cram, Kasman et al. 1998)

- *Medial hamstrings*: 36% of the distance from the ischial tuberoscity to the medial side of the popliteus cavity, starting from the ischial tuberosity (Rainoldi, Melchiorri et al. 2004)

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

- *Vastus medialis oblique*: 20% of the distance from the ASIS to the medial joint space, starting from the joint line (Ericson, Nisell et al. 1985)

- *Vastus lateralis*: 25% of the distance from the ASIS to the lateral joint space, starting from the joint line (Ericson, Nisell et al. 1985)

- *Medial head of gastrocnemius*: 50% of the distance from the medial side of the popliteus cavity to the medial side of the Achilles tendon insertion, starting from the Achilles tendon insertion (Rainoldi, Melchiorri et al. 2004)

Each electrode was placed parallel to the orientation of the muscle fibers; one reference electrode was placed over the anteromedial portion of the proximal tibia. Electrode placement was confirmed with manual muscle testing of each muscle and observation of the muscle activity on an oscilloscope. The electrodes and leads were secured with clear, plastic surgical tape. Each respective muscle group (hip extensors, hip abductors, hip adductors, hamstrings, quadriceps, and plantar flexors) then underwent testing for maximal voluntary isometric contraction (MVIC). Three, 5 second isometric holds, with one minute rest between trials. The MVIC data were used to normalize all EMG activation amplitude data. This was done by dividing the peak MVIC activation averaged over a one second window, by the average EMG activation during the descent phase of the single leg squat. All EMG data were collected at 1000 Hz. The following positions were used for MVIC testing:

- *Hip extensors*: The participant was placed in a prone position with the dominant leg flexed at the knee to 90^0 while he/she attempted to raise his/her thigh off of the table against the downward resistance of the researcher (Kendall FP, McCreary EK et al. 1993). (Figure 3.13)

- *Hip adductors*: The participant was placed in a side lying position on the side of the dominant leg with the nondominant leg flexed at the knee and hip, so that the sole of the participant's nondominant foot could be placed on the testing table in front of the

dominant leg. The participant attempted to adduct his/her hip against the researcher's downward resistance (Kendall FP, McCreary EK et al. 1993). (Figure 3.14)

- *Hip abductors*: The participant was placed in a side lying position on the side of the nondominant leg and he/she attempted to abduct his/her leg against the researcher's downward resistance (Kendall FP, McCreary EK et al. 1993). (Figure 3.15)

- *Hamstrings*: The participant was placed in a prone position with the dominant leg flexed at the knee to 90^0 and the nondominant leg lying flat on the table. He/she attempted to flex his/her knee against the researcher's resistance. (Anderson, Hall et al. 2005). (Figure 3.16)

- *Quadriceps*: The participant was placed in a seated position with both of his/her legs extending off of the table and flexed at the knee to 90^0 and his/her hands crossed across the chest. The participant attempt to extend his/her knee against the researcher's downward resistance (Anderson, Hall et al. 2005). (Figure 3.17)

- *Plantar flexors*: The participant was placed in a prone position with both of his/her legs fully extended, he/she attempted to plantar flex his/her foot against the researcher's resistance (Kendall FP, McCreary EK et al. 1993). (Figure 3.18)

Motion Analysis

Each subject was fitted with three electromagnetic sensors placed over the sacrum, lateral aspect of the thigh, and the anteromedial aspect of the proximal tibia. All sensors were placed over the area of least muscle mass and secured with double-sided tape, prewrap, and athletic tape. Six additional boney landmarks were digitized so joint centers for the ankle and knee were calculated as the midpoints of the respective points based on the position of the medial and lateral malleoli and medial and lateral femoral condyles,

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respectively; the Bell method was used to approximate the hip joint centers (Bell, Pedersen et al. 1990). Three-dimensional coordinate data were collected at a sampling rate of 100 Hz.

Single Leg Squat Task

The subject completed the same procedures for the single leg squat tasks as they did during the screening session. A trial was deemed successful if the participant: 1) maintained proper testing position throughout the entire motion; 2) squatted until the gluteals came in contact with the mechanical block; 3) completed the task at the appropriate rate; 4) the participant did not touch down with the nondominant foot; 5) did not touch the legs together; 6) maintained the heel in contact with the ground and; 7) the task was completed in a fluid motion. The participants were given as many practice trials as needed to perform the task successfully. EMG and motion analysis data were collected simultaneously for five successful SLSs. The participant completed as many trials as necessary until five successful trials were recorded.

Data Processing and Reduction

The Motion Monitor Software (Innovative Sports Training, Inc, Chicago, IL) was used to control both the Motion Star and EMG systems. All kinematic data were filtered using a fourth-order low-pass Butterworth filter at 14.5 Hz. EMG data were passively demeaned, bandpass (10-350 Hz) and notch (59.5-60.5 Hz) filtered, and smoothed using a 25 ms root mean squared sliding window function. Kinematic and EMG data were exported to and reduced using a custom Matlab program (Math Works, Natick, MA).

EMG and kinematic data were recorded during the descent phase of the single leg squat (the time from the start of the trial to peak knee flexion). Three-dimensional joint angles were also recorded at the participant's initial position at the start of the trial (start

position: the participant stood on his/her dominant leg and the nondominant knee was flexed to 90^0 and the hip to 45^0 with the hands on the hips), the point when the participant reached peak knee flexion, and once the participant returned to his/her initial position (end position). EMG and kinematic data were averaged over the five trials of the single leg squat and range of motion data were averaged from the three trials of each passive range of motion measurement.

Statistical Analyses

PASW Statistics for Windows software (version 18.0, SPSS Inc, Chicago, IL) was used to run three separate multivariate analysis of variance (MANOVAs): one MANOVA was run comparing the peak EMG amplitude of the valgus group to the mean peak EMG amplitude of the control group for each of the eight muscles examined in this study; one MANOVA was run comparing the EMG co-activation ratios of gluteus medius to hip adductors and gluteus maximus to hip adductors; one MANOVA was run comparing the comparing the mean passive range of motion measurements of the valgus group to the mean passive range of motion measurements of the control group, for each one of the ten muscle groups examined in this study. Appropriate post hoc one-way between subjects ANOVAs were run for significant MANAOVAs. Statistical significance was set at $\alpha < 0.05$.

RESULTS

Height, Weight, and Age

Means, standard deviations, and 95% confidence intervals for all height, weight, and age measures are presented in Table 4.1. Three separate independent-samples t-tests were utilized to compare the means for each measurement between the control and valgus groups.

No significant differences were observed between height (T_{38} = -.250, P=.804), weight (T_{38} = -.184, P=.855), or age (T_{38} = .000, P=1.00) between the control and valgus groups.

Passive Range of Motion (PROM) Measurements

Means, standard deviations, and 95% confidence intervals for all passive range of motion measures are presented in Table 4.2. A multivariate analysis of variance (MANOVA) test was utilized to compare the mean PROM dependent variables between the control and valgus groups. A significant multivariate main effect for group was observed for PROM measurements (Wilks' Lambda = .555, F(10, 29) P=.038, η^2 = .83). Follow up one-way ANOVAs revealed significant differences in dorsiflexion range of motion with knee straight ($F_{1,38}$ = 4.203, P=.047), dorsiflexion range of motion with knee flexed ($F_{1,38}$ = 4.857, P=.034), and posterior talar glide motion ($F_{1,38}$ = 7.040, P=.012) between the control and valgus groups. Specifically, individuals in the valgus group displayed significantly lesser dorsiflexion PROM with knee straight (5.5±5.4, control = 8.8±4.7), lesser dorsiflexion PROM with knee flexed (9.5±6.2, control = 14.2±7.3) and greater posterior talar glide (29.8±4.8, control = 25.7±5.0) in comparison to the control group. No other significant differences were observed.

Electromyography Measurements

Means, standard deviations, and 95% confidence intervals for all EMG measures are presented in Table 4.3. A MANOVA was run comparing the mean normalized EMG activation between the control and valgus groups. No significant differences were observed between the normalized EMG activation of the two groups (Wilk's Lambda = .742, F(10, 29) P=.280, η^2 = .49). While not significant, it is worth noting the group differences in hip adductor muscle activation was trending toward significance ($F_{1,38}$ = 3.059, P=.089).

Specifically, individuals in the valgus group tended to display greater hip adductor EMG activation (20.1±14.0, control = 13.3±9.8). One valgus subject's EMG data were unable to be used in the MANOVA analysis due to abnormal recording of the hamstring EMG activity.

Electromyography Co-Activation Ratios

Means, standard deviations, and 95% confidence intervals for calculated co-activation ratios presented in Table 4.4. Muscle co-activation ratios were calculated for gluteus medius activation and hip adductor muscle activation by dividing the mean gluteus medius activity by the mean hip adductor activity (GMed : Hip Add). In addition, co-activation ratios were calculated for gluteus maximus activity and hip adductor muscle activation by dividing the mean gluteus maximus activity by the mean hip adductor activation (GMax : Hip Add). A ratio resulting in 1.0 would indicate completely balanced muscular activation; ratios resulting in values greater than 1.0 indicate greater activation of the muscle in the numerator (GMed and GMax) compared to the muscle in the denominator (Hip Add). A MANOVA was run comparing the ratio of gluteus medius EMG activation to hip adductor EMG activation between the control and valgus groups and the ratio of gluteus maximus EMG activation to hip adductor EMG activation between groups. A significant multivariate main effect for group was observed for the co-activation ratios (Wilks' Lambda = .822, F(10,29) *P*=.027, η^2 = .68). Follow-up one-way ANOVAs revealed significant differences between the ratio of gluteus medius EMG activation to hip adductor EMG activation ($F_{1,38}$ = 5.187, *P*=.028) and the ratio of gluteus maximus EMG activation to hip adductor EMG activation ($F_{1,38}$ = 8.201, *P*=.007) between the control and valgus groups. Specifically, individuals in the valgus group displayed significantly lesser gluteus medius EMG activation to hip adductor EMG activation ratio (2.4±1.1, control = 4.5±3.9); indicating greater hip adductor EMG activation

compared to gluteus medius EMG activation in the valgus group when compared to the control group. The valgus group also had significantly lesser gluteus maximus EMG activation to hip adductor EMG activation ratio (1.1±.62, control = 2.4±1.8) compared to the control group; greater hip adductor EMG activation compared to gluteus maximus EMG activation in the valgus group when compared to the control group.

Power and Effect Size

Observed power (range, .050-.797) and effect size (range, .01-1.07) for each PROM and EMG measure were calculated and are presented in table 4.5. This information will be discussed to describe the clinical significance of these findings

DISCUSSION

To our knowledge, this is the first study to compare hip, knee, and ankle passive range of motion (ROM) measurements, and hip, thigh, and lower leg muscle activation between individuals presenting with medial knee displacement (MKD group) to those who do not (control group) during a single leg squat task. In summary, our findings revealed that dorsiflexion ROM measurements with the knee straight and flexed were significantly lesser and posterior talar glide was significantly greater in the MKD group compared to the control group. However, no other ROM measurements (hip internal rotation, hip external rotation, hip abduction, hip extension, knee extension, femoral anteversion) were different between group, which suggests that ROM differences may be isolated to ankle dorsiflexion. Muscle activation amplitude of the gluteus medius, gluteus maximus, hip adductors, vastus medialis, vastus lateralis, medial hamstrings, biceps femoris, and medial gastrocnemius were also not significantly different between the two groups. However, co-activation ratios involving the hip adductors and gluteal musculature were different between groups. Specifically, we

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calculated the co-activation ratio between gluteus medius activation and hip adductor activation (GMed : Hip Add) and gluteus maximus activation and hip adductor activation (GMax : Hip Add). The GMed : Hip Add and GMax : Hip Add ratios were both significantly lesser in the MKD group compared to the control group. Decreased co-activation ratios in the MKD group indicates these individuals use a more hip adductor dominant activation strategy compared to control subjects. Our combined results suggest the combination of lesser ankle dorsiflexion ROM with altered hip adductor and gluteal musculature co-activation may contribute to dynamic knee valgus during a single leg squat task; each of these findings will be discussed in greater detail throughout the subsequent paragraphs.

Lesser ankle dorsiflexion ROM in the MKD group supports our hypotheses. Based on these findings we believe decreased dorsiflexion ROM may be a large contributor to MKD during functional tasks. The MKD group was observed to have 37.5% and 33.1% less dorsiflexion ROM with the knee straight (effect size = 0.65) and flexed (effect size = 0.70), respectively. These values represnt moderate to large effect sizes and further indicate the importance of these differences. Similar findings have been reported when observing ROM differences between a MKD group and a control group who maintained a neutral knee position, in the frontal plane, during a double leg squat (Vesci, Padua et al. 2007; Bell, Padua et al. 2008). Limited dorsiflexion has been proposed to contribute to excessive rearfoot pronation and, in turn, result in compensatory increases in lower extremity internal rotation (DiGiovanni and Langer 2007). Greater lower extremity internal rotation may contribute to dynamic knee valgus (Hollis, Takai et al. 1991). The relationship between decreased dorsiflexion and increased knee valgus during dynamic tasks is further supported by Cortes

et al. who reported that subjects had significantly greater knee valgus angle at initial contact and decreased dorsiflexion motion after landing when they performed rearfoot landings compared to self-preferred landing styles (Cortes, Onate et al. 2007). Based on these combined findings it appears that restricted ankle dorsiflexion ROM may be an important factor contributing to MKD across a variety of functional tasks. These findings may have important implications in the design of exercise programs aimed at decreasing MKD.

Decreased ankle dorsiflexion ROM may be due to decreased flexibility of the gastrocnemius/soleus complex and/or restricted posterior talar glide on the tibia (Denegar, Hertel et al. 2002). Posterior glide of the talus within the talocrural joint is a necessary accessory motion to allow for full ankle dorsiflexion ROM. Previous research investigating subjects with chronic ankle instability has revealed decreased ankle dorsiflexion ROM and posterior talar glide in these individuals (Vicenzino, Branjerdporn et al. 2006). We originally hypothesized decreased ankle dorsiflexion ROM in the MKD group may be due to restricted posterior talar glide motion. However, this hypothesis was not supported as MKD subjects demonstrated greater posterior talar glide compared to the control subjects (effect size = 0.84). This finding suggests that decreased ankle dorsiflexion ROM in the MKD group was most likely due to decreased flexibility of the gastrocnemius/soleus complex and not restricted posterior talar glide. The sensitivity of the posterior talar glide test has been questioned based on research demonstrating weak associations between the posterior talar glide test with open and closed kinetic chain measures of ankle dorsiflexion ROM (Cosby and Hertel 2011). Thus, future research is needed to better understand the underlying mechanism (decreased muscle flexibility or restricted posterior talar glide) contributing to decreased dorsiflexion ROM in the MKD subjects.

There were no other observed significant differences in ROM measurements between groups. Greater hip internal rotation ROM (Clark and Lucett 2004; Hirth 2007) and femoral anteversion (Nguyen and Shultz 2007) have been suggested to lead to increased femoral internal rotation during dynamic tasks, which could contribute to MKD. Likewise, tightness of the hip adductors and internal rotators has been theorized to result in greater femoral adduction and internal rotation (Clark and Lucett 2004; Hirth 2007), again contributing to dynamic knee valgus. Tyler et al. (Tyler, Nicholas et al. 2006) has proposed that the factors contributing to iliotibial band and hip flexor tightness may cause the pelvis to tilt anteriorly and result in internal rotation the femur (Tyler, Nicholas et al. 2006). Greater hamstring flexibility, resulting in lesser activation of the hamstrings, may decrease knee stiffness (Boden, Griffin et al. 2000; Zeller, McCrory et al. 2003) and increase the likelihood of greater MKD during activity. However, our findings indicate the flexibility of these muscle groups did not contribute to the presence of MKD in this study.

Our findings suggest that restricted dorsiflexion ROM appears to be the key factor in predicting MKD during the single leg squat. In our study each participant stood so that his/her foot was fixed on the ground with the toes pointing straight ahead, and we assured the heel remained in contact with the ground throughout each trial. As the subject lowered his/her body to the required 60^0 of knee flexion, dorsiflexion also had to occur at the ankle joint. Limited dorsiflexion ROM would inhibit the tibia from moving forward over the foot and may have caused the subject to compensate for this lack of motion. We speculate that MKD subjects compensated for a lack of sagittal plane ankle motion by increasing frontal and/or transverse plane motion at the foot and up through the kinetic chain. Individuals may have compensated by going into more pronation of the foot, eversion of the talus, and

internal rotation of the tibia (DiGiovanni and Langer 2007) thus creating the visual appearance of medial knee displacement. Future research investigating the three-dimensional kinematics of the foot and lower leg is needed to better understand if these compensatory motions actually do occur in those individuals displaying MKD.

We observed no statistically significant differences in EMG activation between groups for all muscles investigated. These findings are in agreement with comparable research, which has demonstrated few differences in EMG activity during single leg squat tasks; however, these past studies examined EMG differences between sexes and not group assignment based on single leg squat performance. Previous research, looking at differences in muscle activation between sexes during the single leg squat, has shown greater activation of the rectus femoris (Zeller, McCrory et al. 2003) and lesser activation of the gluteus medius (Hart, Garrison et al. 2007) in females, both of which are believed to contribute to increased risk of noncontact ACL injury. Padua et al. reported muscle activation amplitude differences between a group visually displaying MKD and a control group that did not during a double leg squat task. These authors reported 34% greater hip adductor muscle activity in the group displaying MKD (Padua, Bell et al. In review). Similar to this finding, our current study showed the MKD group to have 34% greater EMG activity of the hip adductors during the descent phase of the single leg squat; this finding was not statistically significant, but was trending toward it ($P = .089$). In addition, there were no differences found between groups for the activation of the gluteus medius and gluteus maximus muscles, also similar to Padau et al. (Vesci, Padua et al. 2007; Padua, Bell et al. In review).

Therefore, it is proposed that the relative co-activation between the gluteus medius and gluteus maximus with the hip may contribute to MKD. Increased hip adductor activity

which is not offset by associated increases in gluteus medius and gluteus maximus activation may allow for the femur to be pulled into a more adducted and internally rotated position (Padua, Bell et al. In review). This is supported by our calculated co-activation ratios between the gluteus medius and the hip adductors (GMed : Hip Add) and the gluteus maximus and the hip adductors (GMax : Hip Add), both of which revealed significant differences between groups that are accompanied by large effect sizes. The co-activation ratio of GMed : Hip Add for the MKD group was 2.4 while the control group's ratio was 4.5 (effect size = 0.84); similarly, the ratios for the MKD and control groups for the GMax : Hip Add co-activation ratio were 1.1 and 2.4, respectively (effect size = 1.07). The co-activation ratios were calculated by dividing the gluteal muscle activation (GMed or GMax) by Hip Add activation. Larger co-activation ratios indicate that the GMed or GMax were more active relative to the Hip Add. Conversely, smaller co-activation ratios indicate greater reliance on the Hip Add muscles. Our findings indicate the MKD group places greater reliance on their Hip Add musculature compared to the control group. It is generally thought that MKD may be caused by decreased gluteal muscle strength (Claiborne, Armstrong et al. 2006) or activation (Hart, Garrison et al. 2007). Our findings may help refine this current theory and suggest that MKD may be caused by greater reliance on the Hip Add muscles rather than weakness or decreased activation of the gluteal musculature. We believe that increased hip adductor relative to GMed and GMax activation played a role in facilitating visual MKD during the single leg squat.

There were no differences observed in hip or knee ROM measurements, but there were differences in ankle dorsiflexion ROM. We therefore propose that the imbalance we observed in hip adductor to gluteal activation stems from a neuromuscular compensation as a

result of the decreased ankle dorsiflexion ROM. Decreased ankle dorsiflexion ROM creates an abnormal axis of rotation of the tibia on the talus, resulting from altered arthokinematics which limit roll and glide between the joint surfaces. This abnormal rotation applies abnormal stresses on the tissues which have been suggested to produce altered proprioceptive input, which in turn causes the motor control system to adapt (Denegar and Miller 2002). We propose one such altered motor control response is using more frontal and transverse plane motion when sagittal plane motion is restricted, resulting in the leg being pulled actively inward. One possible neuromuscular mechanism utilized to achieve this is altering the adductor and gluteal co-activation ratios.

Our findings of lesser dorsiflexion ROM measurements and smaller co-activation ratios as potential factors contributing to MKD may have important implications on injury prevention and rehabilitation programs aimed at decreasing MKD. Individuals displaying lesser dorsiflexion range of motion may benefit from increasing gastrocnemius and soleus flexibility. Also, the clinician could utilize an inhibition technique, such as self-myofascial release (ie. foam rolling), to decrease muscle spindle activity and allow the muscle to relax and be further stretched (Hirth 2007). The clinician could also apply a similar treatment to improve the GMed : Hip Add and GMax : Hip Add ratios by inhibiting the hip adductors. In addition the clinician could utilize rehabilitation exercises focused on increasing gluteus medius and gluteus maximus muscle activation and improving neuromuscular control.

Limitations

The following limitations should be considered when interpreting the findings of our study. First, our findings are limited to a single leg squatting task as we did not incorporate other functional tasks in our investigation of range of motion and muscle activation. Future

research should look at whether findings carryover to when individuals perform more challenging dynamic tasks (ie. jump-landing or cutting maneuvers). Also, our findings are limited to healthy, physically active individuals who display visual MKD during a single leg squat and those who did not; therefore, they may not be applicable to an injured population. We cannot speculate if these individuals would display knee MKD during other tasks. In addition, other lower extremity muscles not investigated in this study could be involved in dynamic control during a single leg squat. We are also only able to speculate if MKD during a single leg squat is indicative of increased injury risk during physical activity.

Inherent limitations exist with the use of surface EMG. The assumption was made based on previous literature that EMG signal amplitudes represent levels of muscle activity. Crosstalk may occur with the placement of the EMG surface electrodes on the skin and may not give a true reading of the underlying muscle activity. However, we minimized the potential for error by using standard methods of applying the electrodes, sufficiently securing the electrodes to prevent movement, and checking the output of the electrodes prior to data collection to ensure proper placement. Finally, interpretation of our results was based on EMG signals normalized to maximal isometric voluntary activity (MVIC). Another assumption was made that all participants gave their maximal effort during the MVIC measurements and during the single leg squat; this would affect the normalized percentages used during the statistical analyses.

Another potential limitation of our study is that our measure of posterior talar glide was dependent on the investigator's ability to subjectively determine subtalar neutral position and the end feel/restriction in motion as the knee was moved into flexion. However, the investigator responsible for all ROM measurements established himself to have good

reliability and precision with this measure (ICC = .93, SEM = 1.2^0); therefore, we do not believe this limitation was a major issue with data collection. Future research should look at a more sensitive measure of quantifying restricted posterior talar glide as a possible factor limiting dorsiflexion ROM. Use of an ankle arthrometer to quantify posterior talar displacement and stiffness has been described in previous literature and may be a good tool for future research investigating factors associated with MKD.

Conclusion

In conclusion, our findings indicate that dorsiflexion range of motion measurements were lesser in subjects displaying MKD compared to those who did not. We believe this limited dorsiflexion may result in compensatory movements in the ankle and lower leg, resulting in foot pronation and tibial internal rotation. Greater levels of hip adductor activity without an associated increase in gluteus medius and/or gluteus maximus activity may increase femoral adduction and internal rotation; potentially increasing MKD during dynamic tasks (Padua, Bell et al. In review). MKD is suggested to be a biomechanical factor associated with anterior cruciate ligament injury, medial collateral ligament injury, and patellofemoral pain syndrome. Rehabilitation and injury prevention programs that increase dorsiflexion, decrease hip adductor activity, and increase hip abductor and external rotator activity may potentially decrease the incidence of these injuries.

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