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This research investigated the effects of static lower extremity posture on hip strength, and then examined their collective influence on knee and hip joint kinematics during a single leg squat in males and females. Thirty one healthy males and 31 healthy females, predominantly college students, between the ages of 18 and 35 participated in a single data collection session during which six lower extremity posture characteristics were measured, followed by measurement of hip abduction and extension strength and concluded with neuromuscular and kinematic analysis of the hip and knee during a single leg squat. Hip torque was normalized to body mass and electromyographic data were normalized to maximum voluntary isometric contractions. Five single leg squats were performed on the dominant stance limb to a depth of 60º of knee flexion. Path analysis, implemented by structural equation modeling, was used to examine whether greater lower extremity posture characteristics predicted decreased hip torque and whether greater lower extremity posture characteristics and decreased hip torque collectively predicted greater dynamic valgus knee motion (increased hip adduction and internal rotation, and knee external rotation and valgus). Separate multivariate analyses of variance determined whether females and males differed on measures of lower extremity posture, hip strength, and total hip and knee motion during the single leg squat. The findings were that greater hip anteversion predicted decreased hip abduction torque, and greater tibiofemoral angle predicted decreased hip extension torques. Direct relationships were noted between greater hip anteversion and genu recurvatum with greater knee

external rotation, and between greater navicular drop and hip anteversion with greater hip internal rotation during the single leg squat. Furthermore, decreased hip abduction torque predicted greater knee external rotation while decreased hip extension torque predicted greater knee valgus during a single leg squat. Hence, it was concluded that greater lower extremity posture characteristics predicted decreased posterio-lateral hip strength, and collectively, greater lower extremity posture characteristics and decrease posterio-lateral hip strength predicted greater functional valgus collapse during the single leg squat.

EFFECTS OF LOWER EXTREMITY POSTURE ON HIP STRENGTH AND THEIR INFLUENCE ON LOWER EXTREMITY MOTION DURING A SINGLE LEG SQUAT

by

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Approved by

____
Committee Chair

To my Brothers and Sisters,

You all have always supported me throughout all these years.

To Uncle Jim,

Your generosity and guidance has provided me with endless opportunities.

To my Mother,

Without your amazing courage and sacrifice this achievement would never have been possible.

To my wife Leslie,

Your unconditional love, constant support, and unyielding strength made this all possible. I could not have accomplished this without you.

APPROVAL PAGE

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TABLE OF CONTENTS

Page

LIST OF TABLES ix

LIST OF FIGURES x

CHAPTER

I. INTRODUCTION 1

Statement of the Problem 4
Objective and Hypotheses 5
Independent Variables 6
Dependent Variables 6
Suppressor Variables 7
Limitations and Assumptions 7
Delimitations 8
Operational Definitions 9

II. REVIEW OF LITERATURE 11

Posterio-lateral Hip Musculature 11
Anatomy of the Posterio-lateral Hip Musculature 12
Function of the Posterio-lateral Hip Musculature 13
Posterio-lateral Hip Musculature and Lower Extremity Injury 14
Sex Differences in Posterio-lateral Hip Musculature 16
Lower Extremity Posture and Hip Function 17
Sex Differences in Lower Extremity Posture 20
Pelvic Angle 21
Hip Anteversion 22
Tibiofemoral Angle 24
Quadriceps Angle 26
Genu Recurvatum 28
Pronation 29
Relationship Among Lower Extremity Posture Characteristics 31
Valgus Posture 33
Pronated Posture 35
Clinical Implications 36

Single Leg Squat 38
Summary 39

III. METHODS 40

Subjects 41
Instrumentation 42
Procedures 42
Measurement of LEP 43
Measurement of Hip Torque and MVIC Myoelectrical Signals 45
Measurement of Joint Motion during the Single Leg Squat 49
Data Reduction 52
Statistical Analyses 54

IV. RESULTS 55

Relationships Between LEP, Hip Abduction Torque, and Functional Valgus Collapse 58
Hypothesis 1a: LEP Predicting Hip Abduction Torque 63
Hypothesis 1b: LEP and Hip Abduction Torque Predicting Functional Valgus Collapse 63
Summary of Results Specific to Hypotheses 1a and 1b 64
Relationships Between LEP, Hip Extension Torque, and Functional Valgus Collapse 65
Hypothesis 2a: LEP Predicting Hip Extension Torque 69
Hypothesis 2b: LEP and Hip Extension Torque Predicting Functional Valgus Collapse 69
Summary of Results Specific to Hypotheses 2a and 2b 71
Hypothesis 3: Sex Differences in LEP, Hip Strength and Functional Valgus Collapse 72

V. DISCUSSION 76

Effects of Static Lower Extremity Posture on Hip Strength 77
Hip Anteversion and Hip Abduction Torque 78
Tibiofemoral Angle and Hip Extension Torque 79
Summary 80
Effects of LEP and Hip Strength on Functional Valgus Collapse 81
Direct Effects of LEP on Functional Valgus Collapse 82
Direct Effect of Hip Strength on Functional Valgus Collapse 83
Indirect Effects of LEP on Functional Valgus Collapse 84
Direct and Indirect Effects of Hip Anteversion on Functional Valgus Collapse 84
Relationship between Hip Strength and Activation 86

Sex Differences in Potential Risk Factors for ACL Injury 88
Sex Differences in Lower Extremity Posture 89
Sex Differences in Hip Strength 91
Sex Differences in Functional Valgus Collapse 93
Summary 95
Clinical Implications of Findings.. 97

REFERENCES 100

APPENDIX A. INSTITUTIONAL REVIEW BOARD CONSENT FORM 113

APPENDIX B. ACTIVITY RATING SCALE 115

APPENDIX C_1. PELVIC ANGLE HISTOGRAM 116

APPENDIX C_2. HIP ANTEVERSION HISTOGRAM 117

APPENDIX C_3. QUADRICEPS ANGLE HISTOGRAM 118

APPENDIX C_4. TIBIOFEMORAL ANGLE HISTOGRAM.. 119

APPENDIX C_5. GENU RECURVATUM HISTOGRAM.. 120

APPENDIX C_6. NAVICULAR DROP HISTOGRAM.. 121

APPENDIX C_7. HIP ABDUCTION PEAK TORQUE HISTOGRAM 122

APPENDIX C_8. HIP EXTENSION PEAK TORQUE HISTOGRAM 123

APPENDIX C_9. HIP ADDUCTION EXCURSION HISTOGRAM............................. 124

APPENDIX C_{10}. HIP INTERNAL ROTATION EXCURSION HISTOGRAM.......... 125

APPENDIX C_{11}. KNEE VALGUS EXCURSION HISTOGRAM 126

APPENDIX C_{12}. KNEE EXTERNAL ROTATION EXCURSION HISTOGRAM 127

APPENDIX D_1. LOWER EXTREMITY POSTURE BOX PLOTS BY SEX............. 128

APPENDIX D_2. HIP PEAK TORQUE BOX PLOTS BY SEX.................................. 129

APPENDIX D_3. JOINT EXCURSION BOX PLOTS BY SEX.................................. 130

APPENDIX E. SUBJECT DEMOGRAPHICS, LEP AND HIP TORQUE 131

APPENDIX F. RAW DATA FOR JOINT EXCURSIONS DURING SINGLE LEG SQUAT 134

APPENDIX G. RAW DATA FOR SEMG OF THE POSTERIO-LATERAL HIP 136

LIST OF TABLES

Page

Table 1. Reliability of LEP Measurements 45

Table 2. Previously Established Reliability of MVIC Myoelectrical Signals and Hip Torque 48

Table 3. Previously Established Reliability of MVIC Myoelectrical Signals and Kinematic Measures During a Single Leg Squat 52

Table 4. Descriptive Statistics for Activity Rating Scale, Lower Extremity Posture and Hip Torque 56

Table 5. Descriptive Statistics for Joint Excursion and sEMG 57

Table 6. Correlation Matrix of Predictor and Dependent Variables for Hypotheses 1a and 1b 60

Table 7. Path Coefficients of LEP Predicting Hip Abduction Torque 63

Table 8. Path Coefficients of LEP and Hip Abduction Torque Predicting Knee External Rotation 64

Table 9. Correlation Matrix of Predictor and Dependent Variables for Hypotheses 2a and 2b 67

Table 10. Path Coefficients of LEP Predicting Hip Extension Torque 69

Table 11. Path Coefficients of Hip Extension Torque Predicting Hip Internal Rotation and Knee Valgus 70

Table 12. Descriptive Statistics for Lower Extremity Alignment, Hip Torque and Joint Excursion by Sex 74

Table 13. Results of MANOVA Multivariate Tests 75

LIST OF FIGURES

Page

Figure 1. Moment Arms Acting on the Hip (adapted from Neumann, 1989) 14

Figure 2. Reverse Tailor's Sitting Position..24

Figure 3. Subject Position for Standing MVIC Testing of Hip Abduction 47

Figure 4. Subject Position for MVIC Testing of Hip Extension......................................48

Figure 5. Set-up for Single Leg Squat 50

Figure 6. Sagittal View of Single Leg Squat with Representative sEMG Placement 51

Figure 7. Full Path Model for Dependent Variables Abduction Torque and Functional Valgus Collapse 61

Figure 8. Final Model for Dependent Variables Abduction Torque, Dynamic Valgus, G_{med} Activation 62

Figure 9. Final Model for Dependent Variables Extension Torque, Dynamic Valgus, G_{max} Activation 68

CHAPTER I

INTRODUCTION

"Functional valgus collapse" of the knee, characterized by hip adduction and internal rotation and knee valgus during dynamic activities (Ireland, 1999; Olsen, Myklebust, Engebretsen, & Bahr, 2004) is more commonly observed in females compared to males (Ford, Myer, & Hewett, 2003; Hewett, Myer, & Ford, 2004; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Zeller, McCrory, Kibler, & Uhl, 2003), and has been found to be predictive of ACL injury risk (Hewett et al., 2005). Decreased function of the posterio-lateral hip musculature, in particular the hip abductors and external rotators that are responsible for stabilizing the pelvis and maintaining proper hip and knee alignment, has been postulated as a potential reason for why females have a greater tendency towards this functional valgus collapse (Ford et al., 2003; Griffin et al., 2000; McClay Davis & Ireland, 2003; Zeller et al., 2003). Injury prevention programs have been developed to target the posterio-lateral hip musculature (Hewett, Lindenfeld, Riccobene, & Noyes, 1999), yet clinical screening tools to appropriately identify those with decreased function of the hip musculature are not readily available. Specifically, the single leg squat has been commonly used by clinicians as a functional task to assess hip muscle control, but limited empirical data is available to clearly identify a relationship between hip muscle control and lower extremity motion during the task.

Proper functioning of the hip abductors and external rotators during a single-leg stance is essential to providing proximal stability for lower extremity motion (Kumagai, Naoto, & Higuchi, 1997; Moore, 1992). During single leg weight bearing, the hip abductors must produce a large abduction force to counteract the adduction torque produced from the product of the body weight and its larger moment arm acting at the hip (Neumann, 1989). The gluteus medius is the primary abductor of the hip (Moore, 1992) providing stability in the frontal plane. Although often considered to be a frontal plane muscle, the gluteus medius is also highly active, along with the gluteus maximus, to control rotation in the transverse plane (Earl, 2004; Schmitz, Riemann, & Thompson, 2002). Decreased strength and activation of the posterio-lateral hip would therefore decrease the stability of the hip when it is loaded and result in an inability to maintain a neutral alignment of the hip and knee. This theoretically would result in functional valgus collapse contributing to increased rotational and valgus forces at the knee. Interestingly, it is still unknown whether a relationship exists between decreased neuromuscular function of the hip and increased functional valgus collapse.

While decreased function of the posterio-lateral hip musculature potentially leads to an increase in functional valgus collapse, the underlying causes for this dysfunction and greater prevalence in females have received little attention**.** Sex differences in lower extremity posture, which have also been proposed as an ACL injury risk factor, may alter posterio-lateral hip muscle neuromuscular function and contribute to the differences in dynamic limb alignment between males and females (Ford et al., 2003; Griffin et al., 2000; Hewett et al., 2004; McClay Davis & Ireland, 2003; Zeller et al., 2003). Sex differences in select lower extremity posture characteristics have been reported in pelvic angle, hip anteversion, tibiofemoral angle, quadriceps angle and genu recurvatum (Aglietti, Insall, & Cerulli, 1983; Braten, Terjesen, & Rossvoll, 1992; Guerra, Arnold, & Gajdosik, 1994; Hertel, Dorfman, & Braham, 2004; Horton & Hall, 1989; Hsu, Himeno, Coventry, & Chao, 1990; Nguyen & Shultz, In Press; Prasad, Vettivel, Isaac, Jeyaseelan, & Chandi, 1996; Trimble, Bishop, Buckley, Fields, & Rozea, 2002; Woodland & Francis, 1992). Previous studies that examined the relationship between lower extremity posture characteristics and neuromuscular function suggest differences in alignment can contribute to neuromuscular changes in the hip and lower extremity muscles (Merchant, 1965; Nyland, Kuzemchek, Parks, & Caborn, 2004; Shultz, Carcia, Gansneder, & Perrin, 2006). The limitation with these studies is that only one or select lower extremity posture characteristics were examined. No published studies were found that have examined the relationship between lower extremity posture and hip muscle activation using a comprehensive set of anatomic alignment variables that are sufficiently descriptive of sex differences in lower extremity posture. This is important since one skeletal malalignment may cause compensatory alignment changes at other bony segments resulting in abnormal stress patterns or compensatory motions along the kinetic chain (Gross, 1995; Hruska, 1998; Loudon, Jenkins, & Loudon, 1996). Further research is therefore needed to understand the effects of sex differences in lower extremity posture on hip strength and their influence on lower extremity kinematics and posterio-lateral hip muscle activation during functional weight bearing tasks.

Statement of the Problem

Decreased strength of the posterio-lateral hip musculature may contribute to greater functional valgus collapse of the knee, a position more commonly observed in females compared to males during functional activities (Ford et al., 2003; Hewett et al., 2004; Lephart et al., 2002; Zeller et al., 2003) and one found to be predictive of ACL injury risk (Hewett et al., 2005). Hence, there is a need for clinical assessment tools to identify those with decreased hip muscle strength, which in turn may identify those who are susceptible to joint positions that are known to strain and injure the ACL. Further, identifying factors that influence dynamic hip control during weight bearing activity will provide additional information in the continuing effort to effectively identify those at greater risk for injury and help develop intervention strategies to subsequently reduce the risk of this disabling injury. Therefore, the purpose of this study was to examine the effects of static lower extremity posture on hip strength and their influence on hip and knee kinematics in males and females during a single leg squat, once accounting for activation of the posterio-lateral hip musculature. Specifically, this study examined whether greater relative valgus and pronated lower extremity postures explained decreased hip strength, and whether their collective relationship predicted greater functional valgus collapse (characterized by increased hip adduction and internal rotation, and knee rotation and valgus motion) during a single leg squat. A secondary purpose was to determine whether these characteristics were more pronounced in females compared to males.

Objective and Hypotheses

The objective was to examine the effects of static lower extremity posture on hip strength, and then examine their collective influence on knee and hip joint kinematics during a single leg squat in males and females.

Hypothesis 1a: A static posture characterized by greater hip and knee valgus will explain decreased hip abduction strength.

Hypothesis 1b: Collectively, greater hip and knee valgus and decreased hip abduction strength will predict greater functional valgus collapse (increased hip adduction and internal rotation, and knee external rotation and valgus) during the single leg squat, once accounting for activation of the gluteus medius.

Hypothesis 2a: A static posture characterized by greater hip and knee valgus will explain decreased hip extensor strength.

Hypothesis2b: Collectively, greater hip and knee valgus and decreased hip extension strength will predict greater functional valgus collapse (increased hip adduction and internal rotation, and knee external rotation and valgus) during the single leg squat, once accounting for activation of the gluteus maximus.

Hypothesis 3: Compared to males, females will have 1) a static posture characterized by greater hip and knee valgus, 2) decreased normalized hip strength, and therefore 3) demonstrate greater functional valgus collapse during the single leg squat.

Independent Variables

1. Valgus posture (factor score): Latent variable representing the factor loadings for measurements of anterior pelvic angle, hip anteversion, genu recurvatum, quadriceps angle and tibiofemoral angle.
2. Pronated foot posture (mm): Measurement of navicular drop.

Dependent Variables

1. Hip abduction strength (Nm/kg): Maximum isometric hip abduction torque as measured with an isokinetic dynamometer (standing, hip abducted 5º) and normalized to the subject's body mass.
2. Hip extension strength (Nm/kg): Maximum isometric hip extension torque as measured with an isokinetic dynamometer (supine, hip flexed 90º) and normalized to the subject's body mass.
3. Hip adduction (degrees): Angle of frontal plane excursion of the thigh relative to the sacrum from single leg stance to 60º of knee flexion during a single leg squat.
4. Hip internal rotation (degrees): Angle of transverse plane excursion of the thigh relative to the sacrum from single leg stance to 60º of knee flexion during a single leg squat.
5. Knee external rotation (degrees): Angle of transverse plane excursion of the shank relative to the thigh from single leg stance to 60º of knee flexion during a single leg squat.

6. Knee valgus (degrees): Angle of frontal plane excursion of the shank relative to the thigh from single leg stance to 60º of knee flexion during a single leg squat.
7. Sex: male, female.

Suppressor Variables

1. Gluteus medius muscle activation (%MVIC): The mean RMS amplitude of the gluteus medius from single leg stance to 60º of knee flexion during the single leg squat, normalized to a maximum voluntary isometric contraction.
2. Gluteus maximus muscle activation (%MVIC): The mean RMS amplitude of the gluteus maximus from single leg stance to 60º of knee flexion, normalized to a maximum voluntary isometric contraction.

Limitations and Assumptions

1. Results from this dissertation cannot be generalized to populations other than the college aged individuals utilized, or to tasks other than the single leg squat.
2. All participants provided a maximum effort during testing.
3. The surface electrode placement for the gluteus medius muscle represents myoelectrical activity of the muscle.
4. The surface electrode placement for the gluteus maximus muscle represents myoelectrical activity of the muscle.
5. Surface electromyography amplitude is not analogous to force.

6. Surface electromyography is a reliable and valid method of measuring muscle activity during dynamic activity.
7. Surface electromyography obtained over the electrode placements for each muscle was adequately representative of the muscle as a whole.

Delimitations

1. Only college-aged participants who are healthy with no musculoskeletal injury to either lower extremity for the past 6 months and have not had surgery on either lower extremity participated.
2. All measurements were obtained from the dominant stance leg as determined by the stance leg used to kick a ball.
3. Lower extremity posture characteristics were measured using established clinical measurement methods.
4. Muscle activity and joint kinematics were measured while performing a single leg squat in an upright posture to a depth of 60º of knee flexion.
5. The ensemble average of five consecutive trials represents a participant's single leg squat.
6. Muscle activity for hip abduction was measured via surface electromyography over the gluteus medius.
7. Muscle activity for hip external rotation and extension was measured via surface electromyography over the gluteus maximus.

Operational Definitions

Pelvic angle (PA): The angle (^{o}) formed by a line from the anterior superior iliac spine to the posterior superior iliac spine relative to the horizontal plane as measured by an inclinometer.

Hip anteversion (HA): The angle (^{o}) formed by the shaft of the tibia relative to the vertical plane while in a prone position with the knee flexed to 90° and the greater trochanter positioned in its most lateral position.

Tibiofemoral angle (TFA): The frontal plane angle (^{o}) formed by the anatomical axis of the femur and tibia in a standing position.

Quadriceps angle (QA): The angle (^{o}) formed by a line from the anterior superior iliac spine to the patella center and a line from the patella center to the tibial tuberosity in a standing position.

Genu recurvatum (GR): The sagittal angle (^{o}) formed by the femur and tibia with active contraction of the quadriceps muscle and extension of the knee.

Valgus posture: The scale value representing the relationships among clinical measurements of pelvic angle, hip anteversion, quadriceps angle, tibiofemoral angle and genu recurvatum.

Pronation posture: Clinical measures of navicular drop (mm) representing the difference between the heights of the navicular tubercle measured in subtalar joint neutral and in relaxed stance.

Single leg squat (SLS): A slow descent performed on the dominant leg by flexing the hip, knee and ankle to a depth of 60° of knee flexion with the trunk in an upright position.

Gluteus medius (G_{med}): This muscle is represented by surface electromyography activity obtained at a position one third distance between the iliac crest and the greater trochanter, starting from the greater trochanter, and recorded as the %activation normalized to a maximal voluntary isometric contraction (%MVIC).

Gluteus maximus (G_{max}): This muscle is represented by surface electromyography activity recorded at a position midway between the greater trochanter and the first sacral vertebrae, and recorded as the %activation normalized to a maximal voluntary isometric contraction (%MVIC).

Dominant leg: The stance leg used to kick a ball.

CHAPTER II

REVIEW OF LITERATURE

The goal of this dissertation is to identify the potential effects of static posture on hip strength and their influence on dynamic activation of the hip musculature and joint displacement during a functional clinical assessment. Therefore, this project will quantify the effect of lower extremity posture (LEP) on hip strength and their influence on lower extremity kinematics and activation of the posterio-lateral hip musculature during a single leg squat. This review of literature will provide: 1) a background of the role of the posterio-lateral hip musculature in controlling lower extremity function, 2) current knowledge regarding sex differences in lower extremity posture and their potential influence on posterio-lateral hip activation, and 3) a background regarding the use of the single leg squat as a weight bearing functional task.

Posterio-lateral Hip Musculature

The posterio-lateral hip musculature plays an important role in stabilizing the pelvis and controlling motion of the lower extremity during dynamic activities. Sex differences in hip muscle function may influence joint motion leading to increased lower extremity injury. The following sections will describe the anatomy and function of the posterior-lateral hip musculature, summarize previously reported relationships between hip muscle function and lower extremity injury, and consider the potential influences of sex differences in the posterio-lateral hip musculature that may lead to lower extremity injury.

Anatomy of the Posterio-lateral Hip Musculature

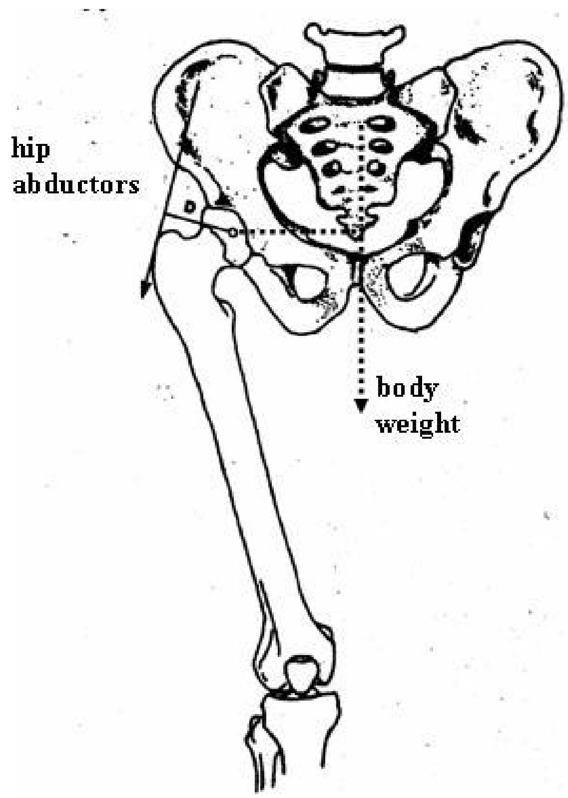
The gluteus medius is the primary abductor of the hip (Inman, 1947; Kumagai et al., 1997; Moore, 1992). The proximal attachment of gluteus medius is along the outer edge of the iliac crest and is fan-shaped spanning the iliac crest from the anterior superior iliac spine to the posterior superior iliac spine. The muscle tapers into a strong tendon and attaches distally on the anterior-superior portion of the greater trochanter. It is defined by three parts, anterior, middle and posterior, that are approximately equal in volume. The anterior and middle fibers run almost vertical while the posterior fibers run horizontal and almost parallel to the neck of the femur (Gottschalk, Kourosh, & Leveau, 1989). Using fine wire EMG, the three segments of the gluteus medius have been found to function in a phasic pattern during submaximal functional activities such as walking, but were found to activate (both onset and duration) more simultaneously during maximal levels of activity such as the support phases of descending stairs (Lyons, Perry, Gronley, Barnes, & Antonelli, 1983; Soderberg & Dostal, 1978).

The gluteus maximus is the largest muscle in the gluteal region and one of the largest muscles in the body (Moore, 1992). The proximal attachments of the gluteus maximus are along the posterior gluteal line of the ilium, dorsal surface of the sacrum and coccyx, and the sacrotuberous ligament. It slopes inferior-laterally at a 45° angle across the ischial tuberosity and attaches distally into the superficial fibers of the iliotibial tract and the gluteal tuberosity of the femur (Kendall, McCreary, & Provance, 1993; Moore, 1992). While the gluteus medius is the primary abductor of the hip, the gluteus maximus functions primarily as an extensor and secondarily as an external rotator of the hip controlling motions in the sagittal and transverse planes (Moore, 1992).

Function of the Posterio-lateral Hip Musculature

Proper functioning of the posterior-lateral hip musculature during single limb weight bearing is essential to providing proximal stability for lower extremity motion. Specifically, the roles of these muscles are to stabilize the pelvis in the frontal and transverse planes to maintain a level pelvis and control rotation at the hip (Earl, 2004; Gottschalk et al., 1989; Inman, 1947; Kumagai et al., 1997; Lyons et al., 1983; Moore, 1992; Schmitz et al., 2002). In the frontal plane, the hip abductors must produce a large abduction torque to counteract the adduction torque produced from the product of the body weight and its larger external moment arm acting at the hip (Neumann, 1989). (Figure 1) Failure to produce the required abduction force is observed as a Trendelenburg posture, with the contralateral pelvis dropping (Neumann, 1989). The important role of the hip abductors in stabilizing the pelvis during single limb function is further illustrated during the midstance phase of gait where activation and force production of the abductors have been observed to be the greatest (Anderson & Pandy, 2003; Neumann, Cook, Sholty, & Sobush, 1992). In the transverse plane, the abductors and external rotators work together to control hip and pelvis motion (Earl, 2004; Schmitz et al., 2002). Hence, weakness or inefficiency of the posterio-lateral hip musculature would decrease stability of the hip when loaded in a single limb weight bearing stance, resulting in an inability to maintain a neutral alignment of the hip and knee.

Figure 1. Moment Arms Acting on the Hip (adapted from Neumann, 1989)



Posterio-lateral Hip Musculature and Lower Extremity Injury

Several retrospective studies have reported decreased strength and activation of the hip abductors in those with low back pain and lower extremity injuries (Beckman & Buchanan, 1995; Brindle, Mattacola, & McCrory, 2003; Bullock-Saxton, 1994; Friel, McLean, Myers, & Caceres, 2006; Ireland, Wilson, Ballantyne, & Davis, 2003; Jaramillo, Worrell, & Ingersoll, 1994; Nadler, Malanga, DePrince, Stitik, & Feinberg, 2000), however, it is unclear if decreased hip function contributed to or resulted from these injuries. Prospective studies that have examined the relationship between function of the posterio-lateral hip musculature and lower extremity injury are limited. Only one study could be found that prospectively examined the relationship between hip strength and lower extremity injury and reported those who experienced injury over the course of the season had significantly less hip abduction and extension strength compared to those who were uninjured (Leetun, Ireland, & Wilson, 2004). The authors suggested that the decreased function of the posterio-lateral hip musculature reduced the ability to stabilize the hip resulting in adduction and rotation of the lower extremity.

"Functional valgus collapse" of the knee, characterized by hip adduction and internal rotation and knee valgus during dynamic movement (Ireland, 1999; Olsen et al., 2004), has been found to be predictive of ACL injury (Hewett et al., 2005). Females, who are at 2-8 times greater risk of suffering an ACL injury (E. Arendt & Dick, 1995; E. A. Arendt, Agel, & Dick, 1999), have consistently been found to demonstrate greater hip adduction and internal rotation and knee valgus when performing functional tasks such as running, landing, jumping, and squatting (Ferber, Davis, & Williams, 2003; Ford et al., 2003; Hewett et al., 2004; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Zeller et al., 2003). In addition, hip and knee internal rotation is also greater in females during running, single-leg landings and forward hopping (Ferber et al., 2003; Lephart et al., 2002). These joint motions contributing to functional valgus collapse have been shown to increase rotational and valgus forces at the knee (Ford et al., 2003; Zeller et al., 2003) and leads to moments acting on the hip that increase the demand of the posterio-lateral hip musculature to stabilize the pelvis and control lower extremity motion (Ferber et al., 2003; McLeish & Charnley, 1970). Sex differences in the function of the hip musculature responsible for stabilizing the pelvis and maintaining proper hip and knee alignment have been postulated as a potential reason for why females have a greater tendency towards this functional valgus collapse (Ford et al., 2003; Griffin et al., 2000; McClay Davis & Ireland, 2003; Zeller et al., 2003).

Sex Differences in Posterio-lateral Hip Musculature

Strength and activation of the posterio-lateral hip musculature appears to be sex dependent. Several studies that examined strength of the hip muscles report females consistently produce a significantly lower amount of force/torque in the direction of hip abduction (Bohannon, 1997; Cahalan, Johnson, Liu, & Chao, 1989; Leetun et al., 2004; Murray & Sepic, 1968), extension (Cahalan et al., 1989) and external rotation (Cahalan et al., 1989; Leetun et al., 2004) compared to males. While females have also been observed to have less gluteus medius (~15-17%) during functional activities; sex differences in gluteus maximus activation are conflicting where greater activation was observed in females during a single leg squat (Zeller et al., 2003) while less activation was observed in females during single leg landings (Zazulak et al., 2005). Small sample sizes and methodological considerations in performing the tasks may explain these contrasting findings. Specifically, trunk motion does not appear to be controlled in these studies, which has been shown to have a direct influence on activation of the hip musculature (Schmitz et al., 2002).

Although limited, these observed sex differences in activation of the posterior-lateral hip function may offer a potential reason for why females may have a greater tendency towards dynamic malalignments leading to functional valgus collapse. However, the underlying causes for this dysfunction and greater prevalence in females have received little attention**.**

<u>Lower Extremity Posture and Hip Function</u>

Research suggests that lower extremity posture, which has also been proposed as an ACL injury risk factor, may alter posterio-lateral hip muscle function and contribute to functional valgus collapse (Ford et al., 2003; Griffin et al., 2000; Hewett et al., 2004; McClay Davis & Ireland, 2003; Zeller et al., 2003). Studies examining the direct influence of lower extremity posture on posterio-lateral hip muscle function are limited, but suggest that differences in lower extremity alignment may contribute to changes in the force and activation of the posterio-lateral hip musculature. For example, increased hip anteversion, which results in increased femoral internal rotation, may alter hip muscle function leading to reduced hip control and increased dynamic lower extremity malalignments during functional activities (Nyland et al., 2004). Using a simulated hip model, an increase in gluteus medius force was necessary to maintain a level pelvis when the femur was internally rotated (as in the case with hip anteversion) compared to a neutral alignment (Merchant, 1965). Further, decreased activation of the gluteus medius as measured by surface electromyography amplitude has been demonstrated in those with increased relative hip anteversion (Nyland et al., 2004). Collectively these findings suggest that individuals who have increased hip anteversion will require increased force production to control the hip and pelvis, yet demonstrate decreased activation, which together may severely reduce frontal and transverse plane hip control during functional activities.

Reasons that may explain the relationship between the posterio-lateral hip musculature and hip anteversion, as well as other lower extremity posture characteristics, are the influence of these characteristics on the moment arms of the muscles. As previously mentioned, in a static single leg stance, the hip abductors must produce a large abduction force to maintain and stabilize a level pelvis. This is because the muscle has a relatively small internal moment arm to counteract the adduction force produced from the product of the body weight and its larger external moment arm acting at the hip (Neumann, 1989). In theory, lower extremity postures that would further decrease the internal moment arms of the posterio-lateral hip musculature would increase the muscular demand to maintain stability of the pelvis. If these same lower extremity postures also reduce the muscular efficiency and force producing capabilities of the muscle, stabilization of the pelvis may become very difficult.

The anterior portion of the gluteus medius has a small internal rotation moment arm while the middle and posterior portions have an abduction and external rotation moment arm when the hip is neutral in the sagittal plane (Delp, Hess, Hungerford, & Jones, 1999; Mansour & Pereira, 1987). As the hip flexes, the origin of the gluteus medius is displaced anteriorly, resulting in the muscle functioning more as an internal rotator of the hip (Delp et al., 1999). Specifically, research has shown that the internal rotation moment arm of the anterior portion of the gluteus medius increases while the abduction and external rotation moment arms of the middle and posterior portions decrease and switch towards internal rotation (Delp et al., 1999; Dostal & Andrews, 1981; Dostal, Soderberg, & Andrews, 1986). Theoretically, an increase in anterior pelvic angle and hip anteversion would position the origin of gluteus medius more anteriorly, resulting in the muscle functioning more as an internal rotator, which in turn may increase the demand of the gluteus maximus to counteract and control internal rotation of the hip and knee. This increased demand of the gluteus maximus resulting from these faulty postures may further compound an already compromised muscle. While the gluteus maximus primarily has an external rotation moment arm when the hip is in neutral, as the hip flexes, the moment arms for portions of the muscle switch to internal rotation while the external rotation moment arms in other portions are reduced (Delp et al., 1999).

Collectively, these findings suggest that lower extremity postures characterized by an anterior pelvic tilt and increased hip internal rotation may decrease strength of the posterio-lateral hip muscles while increasing the demands on the muscles to stabilize the hip and pelvis. In turn, these neuromuscular and biomechanical changes may explain the decreased frontal and transverse plane hip control (i.e. functional valgus collapse) more often observed in females compared to males during functional activities.

Sex Differences in Lower Extremity Posture

Previous literature examining sex differences in lower extremity posture consistently demonstrates greater quadriceps angle (Q-angle) in females (Aglietti et al., 1983; Guerra et al., 1994; Hertel et al., 2004; Horton & Hall, 1989; Hsu et al., 1990; Woodland & Francis, 1992), while limited studies support a sex difference in pelvic angle (Hertel et al., 2004) and genu recurvatum (Trimble et al., 2002). No sex differences have been observed in measures of tibial torsion (Pasciak, Stoll, & Hefti, 1996; Staheli, Corbett, Wyss, & King, 1985) and foot pronation as measured by navicular drop (Beckett, Massie, Bowers, & Stoll, 1992; Hertel et al., 2004; Moul, 1998; Trimble et al., 2002) and rearfoot angle (Astrom & Arvidson, 1995; Sobel, Levitz, Caselli, Brentnall, & Tran, 1999), and sex differences in measures of hip anteversion (Braten et al., 1992; Prasad et al., 1996) and tibiofemoral angle remain unclear (Cooke et al., 1997; Hsu et al., 1990; Tang, Zhu, & Chiu, 2000).

The limitation with many of these studies is that, in most cases, the sample sizes were relative small and only select alignment variables were examined. However, a recently completed study (Nguyen & Shultz, In Press) examining sex differences in a comprehensive set of lower extremity posture characteristics in a relatively large cohort (N=100) provides further empirical evidence that females tend to stand with greater anterior pelvic angle, hip anteversion, knee hyperextension and knee valgus (Q-angle and tibiofemoral angle) compared to males. This study also supports previous work finding no sex differences in the lower legs, ankles or feet with measures of tibial torsion, navicular drop and rearfoot angle. While the reasons contributing to these sex differences are unknown, there is evidence to suggest that the sex differences are developmental in nature, emerging post puberty (Cahuzac, Vardon, & Sales de Gauzy, 1995; Crane, 1959; McDonough, 1984; Salenius & Vankka, 1975; Svenningsen, Apalset, Terjesen, & Anda, 1989; Vankka & Salenius, 1982). The following sections will describe these lower extremity posture characteristics, summarize reported normative values, and consider the potential influences that may lead to excessive angulations and the sex differences reported.

Pelvic Angle

The neutral position of the pelvis is when the anterior superior iliac spines (ASIS) lie in the same transverse plane with one another and the posterior superior iliac spines (PSIS), and are aligned in the same frontal plane as the pubic symphysis (Kendall et al., 1993). A deviation from this neutral position is defined by the orientation of the ASIS relative to the PSIS in the transverse plane. Pelvic angles where the PSIS are above the horizontal in reference to the ASIS are considered an anterior pelvic tilt. This occurs as the ASIS moves anterior and inferiorly and the PSIS moves superiorly. Conversely, pelvic angles where the PSIS are below the horizontal relative to the ASIS are considered a posterior pelvic tilt. This occurs as the ASIS moves posterior and superiorly and the PSIS moves inferiorly (Kisner & Colby, 1996; Norkin & Levangie, 1992; Sanders & Stavrakas, 1981).

Previous reported mean values of pelvic angles on adult subjects range from approximately 9^{o} to 12^{o} of anterior pelvic tilt using various measurement methods

(Alviso, Dong, & Lentell, 1988; Day, Smidt, & Lehmann, 1984; Gajdosik, Simpson, Smith, & DonTigny, 1985; Gilliam, Brunt, MacMillan, Kinard, & Montgomery, 1994; Levine & Whittle, 1996; Nguyen & Shultz, In Press; Shultz, Nguyen et al., 2006). Those that examined sex differences observed 2-4^{o} greater anterior pelvic tilt in females compared to males (Hertel et al., 2004; Nguyen & Shultz, In Press). While reasons to explain a sex difference in pelvic angle are unknown, it has been suggested that an imbalance in the muscles that control the pelvis (i.e. rectus abdominis, erector spinae, gluteal muscles, flexors of the hip, etc.) can contribute to differences in pelvic angle (Hruska, 1998; Kendall et al., 1993). Tightness causing shortening of the erector spinae and hip flexors and/or weakening causing elongation of the abdominals and gluteals has been suggested to increase anterior pelvic tilt. The relationship between pelvic angle and these muscles, specifically the gluteal muscles, may potentially explain the sex difference in pelvic angle as females have been observed to have decreased strength in hip abduction (Bohannon, 1997; Cahalan et al., 1989; Leetun et al., 2004; Murray & Sepic, 1968), extension (Cahalan et al., 1989) and external rotation (Cahalan et al., 1989; Leetun et al., 2004) compared to males.

Hip Anteversion

Torsion of the femur is represented by the angle formed between the axis of the femoral neck and a transverse line through the femoral condyles, also known as the transcondylar axis or plane (Crane, 1959; Norkin & Levangie, 1992). Hip anteversion is defined as the forward projection of the femoral neck from the transcondylar plane. This lower extremity posture characteristic is developmental with age and is greatest at birth (approximately 35° to 40°) gradually decreasing to approximately 12° to 15° in adulthood (Crane, 1959; McDonough, 1984; Svenningsen et al., 1989). Normal mean values reported in healthy adults range from 7^{o} to 18^{o} using both clinical and diagnostic measurement methods (Braten et al., 1992; Jonson & Gross, 1997; Nguyen & Shultz, In Press; Pasciak et al., 1996; Prasad et al., 1996; Reikeras & Bjerkreim, 1982; Schneider et al., 1997; Seber et al., 2000; Shultz, Nguyen et al., 2006). While hip retroversion is used to describe torsion of the femur when the femoral neck is posterior to the transcondylar plane, it is also used to describe hip anteversion that has regressed to the point that it is lower than the normal range (Gulan, Matovinovic, Nemec, Rubinic, & Ravlic-Gulan, 2000).

Increased hip anteversion has been suggested to result from inadequate regression of anteversion from infancy to adulthood (McDonough, 1984). While reasons for a lack of regression are unknown, heredity has been suggested to play role as increased hip anteversion is frequently present in the mother of children with increased hip anteversion (Staheli, 1977). Behavioral factors that increased stress on the medial femoral growth plate through childhood have also been suggested to contribute to excessive hip anteversion. These include sitting in the “reverse tailor’s” position (Figure 2) and frequent in-toe belly sleeping (McDonough, 1984). This lack of developmental regression may contribute to greater hip anteversion in females compared to males reported in some studies (Braten et al., 1992; Nguyen & Shultz, In Press; Prasad et al., 1996). However, it is unknown if females have greater hip anteversion during infancy, or if the regression is slower than males during puberty secondary to an increased prevalence of the behavioral factors mentioned.

Figure 2. Reverse Tailor's Sitting Position

Tibiofemoral Angle

Tibiofemoral angle represents the alignment of the long axis of the tibia, relative to the long axis of the femur, in the frontal plane. The axes of each segment are defined by an anatomical axis and a mechanical axis respective to each segment. Moreland et al. (1987) has described these axes in detail where the anatomical and mechanical axes of the tibia are the same and are represented by a line between the knee joint center and the ankle joint center. The mechanical axis of the femur represents a line from the center of the head of the femur to the knee joint center. The anatomical axis of the femur is a line that qualitatively represents the shaft of the femur through the knee joint center. When the femur and tibia are aligned in a straight line in the frontal plane, this is assumed to be the neutral position. Any deviation from this neutral position is commonly referred in terms of varus ("bow-legged") or valgus ("knock-knees") alignment of the knee.

Due to differences in measurement methods, normal values for tibiofemoral angle are unclear and are dependent on whether the anatomical or mechanical axis of the femur is used for measurement. Longitudinal studies have demonstrated that tibiofemoral angle follows a pattern of development from infancy to adolescence beginning with a varus deformity, progressing to a valgus deformity with the start of bipedal walking, and finally a regression of the valgus deformity (Salenius & Vankka, 1975; Vankka & Salenius, 1982). It is reported that a varus deformity is present in children less than one year old which tends to decrease to where the knee is straight at the approximate one to one and a half years of age. The progression towards valgus continues, becoming most pronounced at around two to three years of age, but then decreases (towards a varus) by the age of six to seven. In the adult population, studies using the anatomical axis report mean values of a valgus alignment (Cahuzac et al., 1995; Hsu et al., 1990; Nguyen & Shultz, In Press; Shultz, Nguyen et al., 2006) while mean values using the mechanical axis indicate a varus alignment (Cooke et al., 1997; Moreland et al., 1987; Tang et al., 2000). The primary reason for this disparity may that the anatomical axis of the femur has a normal valgus angulations of 5-7^{o} relative to the mechanical axis (Oswald, Jakob, Schneider, & Hoogewoud, 1993).

Studies that have examined sex differences in tibiofemoral angle using the anatomical axis of the femur report greater valgus angles in females (Hsu et al., 1990; Nguyen & Shultz, In Press) while those using the mechanical axis of the femur report no sex difference (Cooke et al., 1997; Moreland et al., 1987; Tang et al., 2000). This discrepancy may be attributed to measurements using the mechanical axis of the femur, which does not account for structural abnormalities at the femoral neck that may predispose an individual to knee valgus such as coxa vara (neck-shaft angle <125 ^{o}) (Powers, 2003). While no sex differences are present prior to adolescence (Salenius & Vankka, 1975; Vankka & Salenius, 1982), there is evidence to suggest that sex differences in the rate of development in tibiofemoral angle are a result of changes occurring during the adolescent years of growth (Cahuzac et al., 1995). This is supported by findings that sex differences are present in the decline of valgus alignment through the adolescent years; whereas boys continue to move towards a varus or more neutral alignment with significant decreases in valgus alignment after the age of 13, girls maintain a valgus alignment (Cahuzac et al., 1995). While this may contribute to the greater valgus alignment found in females post puberty, the reasons to explain this difference in rates of development are unknown.

Quadriceps Angle

The quadriceps angle or Q-angle is a clinical measurement that is used to represent the resultant quadriceps muscle force on the patella in the frontal plane (Schulthies, Francis, Fisher, & Van de Graaff, 1995). It represents the angle formed by the vectors for the combined pull of the quadriceps femoris and the patellar tendon (Hungerford & Barry, 1979). Clinically, the Q-angle represents the intersection of a line from the anterior superior iliac spine to the center of the patella and a line from the center of the patella to the tibial tuberosity. A normal angle of 10° has been suggested for measurement of Q-angle with angles greater than 15° considered to be abnormal (American Orthopaedic Association, 1979). However, these values appear to be based on clinical observation and do not account for differences by sex. Normative values by sex reported in the literature range from 8-15° in males and 12-19° in females (Aglietti et al., 1983; Guerra et al., 1994; Horton & Hall, 1989; Hsu et al., 1990; Livingston & Mandigo, 1997; Nguyen & Shultz, In Press; Woodland & Francis, 1992). When these sex differences are considered, angles greater than 15° for men and greater than 20° for women have been suggested as clinically abnormal (Hvid, Andersen, & Schmidt, 1981).

The Q-angle is consistently reported to be greater in females but reasons to explain a sex difference in Q-angle are still unknown. The previous thought of larger Q-angles in females resulting from an increased hip width compared to males has been well disputed (Guerra et al., 1994; Horton & Hall, 1989; Kernozek & Greer, 1993). In fact, in one of the studies reporting no relationship between increased hip width and Q-angle, males were observed to have larger mean hip widths compared to females (Horton & Hall, 1989). More likely, sex differences in Q-angle are a result of sex differences in other lower extremity posture characteristics along the lower extremity that may change the position of the anatomical landmarks of the patella and tibia used in the Q-angle measurement. For example, increased hip anteversion would result in movement of the patella medially relative to the anterior superior iliac spine (Powers, 2003). Further, hip anteversion has been shown to be compensated for by increased external rotation of the tibia which would result in movement of the tibial tuberosity laterally (Hvid & Andersen, 1982). Increased tibiofemoral angle also has the potential to alter the Q-angle, as it would position the tibial tuberosity more laterally.

Genu Recurvatum

Genu recurvatum, or hyperextension of the knee, is defined as sagittal alignment of the lateral midline of the femur and lower leg at the tibofemoral joint beyond the zero position of extension (Kendall et al., 1993). A range of 0°-5° of genu recurvatum has been suggested as normal (Loudon et al., 1996), however, it is unclear whether this applies to both sexes. Studies that report normal values by sex are limited and provide a wide range with mean values for females between 0.2°- 6.1° and -0.3°-3.2° for males (Nguyen & Shultz, In Press; Scerpella, Stayer, & Makhuli, 2005; Trimble et al., 2002). Of these studies, one exclusively examined a collegiate athletic population (Scerpella et al., 2005) which represents the lower range of mean values reported. When this study is excluded, the range of mean normative values in a healthy population fall between 5.8°-6.1° for females and 2.3°-3.2° for males (Nguyen & Shultz, In Press; Trimble et al., 2002). Further, a sex difference was not reported in an athletic population (Scerpella et al., 2005) but was observed in the studies that examined a heterogeneous population (Nguyen & Shultz, In Press; Trimble et al., 2002). This suggests that measures of genu recurvatum may be both sex and population dependent.

As with many of the other alignment variables, the reasons to explain a sex difference in genu recurvatum are unknown. Increased laxity of the ACL has been suggested to contribute to genu recurvatum at the knee (Noyes, Dunworth, Andriacchi, Andrews, & Hewett, 1996) since the ACL is taut when the knee is in full extension (Norkin & Levangie, 1992). Greater anterior laxity of the knee, a motion largely restricted by the ACL, has been reported in females compared to males (Rosene & Fogarty, 1999; Rozzi, Lephart, Gear, & Fu, 1999; Shultz, Kirk, Sander, & Perrin, 2005) and could potentially lead to greater genu recurvatum in females. In addition, sex differences in genu recurvatum may also be related to the increased anterior pelvic tilt found in females (Hertel et al., 2004; Nguyen & Shultz, In Press). An excessive anterior pelvic tilt has been suggested to create a flexion moment at the hip that is counteracted with an extension moment at the knee resulting in hyperextension at the knee joint (Kendall et al., 1993). However, it is unknown if hyperextension is a compensation for greater anterior pelvic angle or if increased pelvic angle is a compensation for greater genu recurvatum resulting from increased anterior knee laxity.

Pronation

Subtalar joint pronation, when measured in a closed kinetic chain, is a combination of calcaneal eversion with adduction and plantar flexion of the talus (Root, Orien, Weed, & Hughes, 1977). Clinically, pronation is commonly examined by measures of navicular drop and rearfoot angle. Navicular drop is commonly defined as the difference between the height of the navicular in subtalar joint neutral and the resting height of the navicular in a relaxed stance (Brody, 1982). Normative values in the adult population using similar measurement methods report mean values ranging from 6-9 mm (Beckett et al., 1992; Evans, Copper, Scharfbillig, Scutter, & Williams, 2003; Nguyen & Shultz, In Press; Picciano, Rowlands, & Worrell, 1993; Shultz, Nguyen et al., 2006; Trimble et al., 2002). A navicular drop of 10 mm has been suggested as normal and values greater than 15 mm considered abnormal (Brody, 1982), however, no quantitative data were reported to support these limits. Others have suggested navicular drop values greater than 13 mm (Beckett et al., 1992) and 10 mm (Mueller, Host, & Norton, 1993) to be abnormal.

Rearfoot angle is formed by the angle of the calcaneus in reference to the lower leg and is often reported as a deviation from the subtalar joint neutral position. Clinically, frontal plane alignment of the rearfoot is often utilized as an indirect measurement of subtalar joint pronation (T. McPoil & Cornwall, 1994). Normative values of rearfoot angle in the adult population using similar measurement methods report mean values ranging from 4º-8º of eversion (Astrom & Arvidson, 1995; Nguyen & Shultz, In Press; Woodford-Rogers, Cyphert, & Denegar, 1994). Rearfoot angles greater than 5° of eversion have been considered to be abnormal as this disturbs the axis of the foot and the normal distribution of pressure (LeLievre, 1970). The literature provides no evidence of differences between males and females in measurement of navicular drop (Beckett et al., 1992; Hertel et al., 2004; Moul, 1998; Nguyen & Shultz, In Press; Trimble et al., 2002) or rearfoot angle (Astrom & Arvidson, 1995; Nguyen & Shultz, In Press; Sobel et al., 1999).

Relationship Among Lower Extremity Posture Characteristics

As previously stated, sex differences in lower extremity posture have been suggested to alter posterio-lateral hip function and are included as one of the risk factors potentially contributing to the increased prevalence of ACL injury in females (Ford et al., 2003; Griffin et al., 2000; Hewett et al., 2004; Hutchinson & Ireland, 1995; Ireland, 1999; Ireland, Gaudette, & Crook, 1997; McClay Davis & Ireland, 2003; Zeller et al., 2003). However, the relationship between lower extremity posture characteristics, and how they collectively contribute to posterio-lateral hip function and ACL injury are not clearly understood. While no single postural characteristic has been reliably associated with an increase rate of ACL injury (Griffin et al., 2006), accounting for all relevant lower extremity posture characteristics may more accurately describe the relationship between posterio-lateral hip function and ACL injury risk since one alignment characteristic may interact with or cause compensations at other bony segments (Gross, 1995; Hruska, 1998; Loudon et al., 1996).

In support of this theory, interactive effects between select lower extremity posture characteristics and neuromuscular activation of the thigh and calf muscles have been observed in response to postural perturbations (Shultz, Carcia et al., 2006). Findings revealed that subjects classified as having above average navicular drop and Q-angle exhibit very different neuromuscular responses in the quadriceps and hamstrings depending on whether one or both of these alignment characteristics were present. Hence, it is plausible that lower extremity posture characteristics may also interact to influence neuromuscular responses in the posterio-lateral hip musculature as well.

Additionally, several studies support an interactive effect between lower extremity posture characteristics in relation to ACL injury risk. While pronation has been the postural characteristic most consistently linked to ACL injury risk (Beckett et al., 1992; Hertel et al., 2004; Loudon et al., 1996; Woodford-Rogers et al., 1994), stronger relationships have been reported when pronation is examined in combination with pelvic tilt (Hertel et al., 2004), genu recurvatum (Loudon et al., 1996) and knee laxity (Woodford-Rogers et al., 1994). The potential for lower extremity posture characteristics to combine and interact to effect neuromuscular function (Shultz, Carcia et al., 2006) and predict the likelihood of suffering an ACL injury (Hertel et al., 2004; Loudon et al., 1996; Woodford-Rogers et al., 1994) reinforces the need to take a comprehensive approach to fully understand the relationships between lower extremity posture, dynamic knee joint function and injury risk. Hence, identifying relationships among lower extremity posture characteristics is a crucial step towards understanding their interactive effects on lower extremity function.

A recently completed study appears to be the only study that provides empirical data identifying relationships among a comprehensive set of lower extremity posture (Nguyen & Shultz, In Press). Using a factor analysis approach, two distinct lower extremity postures were identified in a cohort of 100 subjects, a relative valgus posture and a relative pronated posture. The valgus posture identified positive relationships between greater anterior pelvic angle, hip anteversion, tibiofemoral angle, Q-angle and genu recurvatum. The pronated posture identified positive relationships between greater navicular drop and rearfoot angle. Interestingly, sex differences have been observed in the lower extremity posture characteristics that define the valgus posture (Aglietti et al., 1983; Braten et al., 1992; Guerra et al., 1994; Hertel et al., 2004; Horton & Hall, 1989; Hsu et al., 1990; Livingston & Mandigo, 1997; Nguyen & Shultz, In Press; Prasad et al., 1996; Shultz, Nguyen et al., 2006; Trimble et al., 2002; Woodland & Francis, 1992), but not in those that define the pronated posture (Astrom & Arvidson, 1995; Beckett et al., 1992; Hertel et al., 2004; Moul, 1998; Nguyen & Shultz, In Press; Sobel et al., 1999; Trimble et al., 2002).

Valgus Posture

Based on clinical expertise and observation, excessive anterior tilt of the pelvis is thought to lead to alignment changes in the lower kinetic chain, specifically internal femoral rotation, genu valgus and genu recurvatum (Hruska, 1998; Ireland et al., 1997; Powers, 2003). The relationship between pelvic angle and genu recurvatum is logical as both are sagittal plane alignments where excessive anterior tilt of the pelvis creates a flexion moment at the hip that is counteracted with an extension moment at the knee (Kendall et al., 1993). When measured in weight bearing, internal rotation of the femur has been attributed to increased pelvic angle as a result of a change in the orientation of the acetabulum (Hruska, 1998) in combination with hyperextension of the knee (Kendall et al., 1993).

Femoral internal rotation, which can result functionally from increased anterior pelvic tilt or structurally from increased hip anteversion, can also influence transverse and frontal plane knee angles (i.e. tibiofemoral angle and Q-angle) by changing the spatial orientation of the anatomical landmarks used for these measurements. The most common problem associated with increased hip anteversion is an intoeing gait (Gulan et al., 2000; Staheli, 1977), which results in medial rotation of the patella (Gulan et al., 2000) and internal rotation of the tibia during walking (Staheli, 1977). This compensation would effectively displace the anatomical axes of the femur into adduction and the tibia into abduction, thereby altering the measurement of tibiofemoral angle. In addition, this abnormal gait pattern can also indirectly lead to compensations in other parts of the lower extremity, such as a compensatory external rotation of the tibia on the femur (Fabry, MacEwen, & Shands, 1973). This in turn would position the tibial tuberosity more laterally, resulting in an increased Q-angle. These compensations may explain the significant positive correlation that has been observed between femoral anteversion and Q-angle (Hvid & Andersen, 1982). Similarly, a collective lower extremity posture that includes a combination of greater hip anteversion and knee valgus (movement of the patella medially relative to the ASIS and tibial tubercle), and external tibial rotation (movement the tibial tubercle laterally) would reflect an increase in Q-angle (Hvid & Andersen, 1982; Powers, 2003; Woodland & Francis, 1992).

While it has been proposed that excessive Q-angle may increase the risk of ACL injury, very little research has examined this relationship (Hertel et al., 2004; Loudon et al., 1996). As previously mentioned, greater Q-angle may result from movement of the patella medially and/or movement of the tibial tuberosity laterally with greater tibiofemoral angle and hip anteversion (Hvid & Andersen, 1982; Powers, 2003) (which also may result from increased pelvic angle changing the orientation of the acetabulum

(Hruska, 1998)). Given the identified relationships between these variables, and the potential for any one of these variables to differentially influence the Q-angle, highlights the potential difficulty when independently examining the relationship between Q-angle and lower extremity motion leading to ACL injury risk. This further supports the need to consider the collective influences of lower extremity posture characteristics as a combined posture, rather than as independent alignment variables.

Pronated Posture

Measurement of pronation, a tri-planar movement which involves calcaneal eversion, talar adduction and plantar flexion, and forefoot abduction, (Root et al., 1977) has been commonly performed using both navicular drop (Beckett et al., 1992; Brody, 1982; Trimble et al., 2002; Woodford-Rogers et al., 1994) and rearfoot angle measures (Astrom & Arvidson, 1995; Livingston & Mandigo, 2003; T. G. McPoil & Cornwall, 1996; Smith-Oricchio & Harris, 1990; Sobel et al., 1999). It was not surprising that findings identified a relationship between these variables, as rearfoot position has been found to contribute to the measure of navicular drop (Mueller et al., 1993). Of particular interest and importance is that this pronated posture (i.e. the combination of navicular drop and rearfoot angle) was independent of the valgus posture describing alignment of the hip and knees (Nguyen & Shultz, In Review). This suggests that the alignment differences in the proximal segments of the hips and knees are independent of those in the distal segments of the foot and ankle. Hence, both postural factors should be accounted for when examining the relationship between lower extremity posture, dynamic hip and knee function, and the potential for injury.

Clinical Implications

The combination of the hip and knee lower extremity posture characteristics (pelvic angle, hip anteversion, Q-angle, tibiofemoral angle and genu recurvatum) that define the valgus posture suggest the potential for inward collapse of the knee when increased values are observed in a static posture. While the clinical implications of the increased frontal plane knee angles, as measured statically, are relatively unknown, females have been consistently found to land and cut with greater valgus angles and moments compared to males (Ford et al., 2003; Hewett et al., 2004; Hewett et al., 2005; Lephart et al., 2002; Zeller et al., 2003). Whether this postural factor may in part explain why females have been consistently found to land and cut with greater valgus angles and moments compared to males (Ford et al., 2003; Hewett et al., 2004; Hewett et al., 2005; Lephart et al., 2002; Zeller et al., 2003) or whether this is due to anatomical versus neuromuscular differences, or both, is unknown. However, as previously mentioned there is evidence to support that pelvic angle and hip anteversion may change the moment arms of the posterio-lateral hip muscles (Delp et al., 1999; Dostal & Andrews, 1981; Dostal et al., 1986; Mansour & Pereira, 1987), which are responsible for controlling dynamic motion and alignment of the lower extremity during functional activities (Neumann, 1989; Nyland et al., 2004). Future studies are needed to examine whether this static valgus posture may contribute to sex differences in dynamic hip and knee joint function, and the increased risk of functional valgus collapse and ACL injury in females.

While sex differences have not been observed in measures of navicular drop and rearfoot angle that define the pronated posture, it cannot be ignored that subtalar joint pronation is the alignment factor that has been most consistently linked to ACL injury (Beckett et al., 1992; Hertel et al., 2004; Loudon et al., 1996; Woodford-Rogers et al., 1994). It is also important to note that the predictive strength of this variable is notably greater when examined in combination with other proximal lower extremity alignment variables. As previously described, retrospective studies have found that pronation is a stronger predictor of ACL injury risk when considered in combination with anterior pelvic tilt (Hertel et al., 2004) and genu recurvatum (Loudon et al., 1996), two variables that are related to the internal femoral rotation and valgus posture previously described. Although the valgus and pronated postures are independent of one another, it may be that the interaction of these postures in some way influences the ACL injury equation. Specifically, individuals who demonstrate alignment variables consistent with increased subtalar pronation and increased valgus posture (e.g. excessive anterior pelvic angle, hip anteversion, tibiofemoral angle, Q-angle and genu recurvatum) may combine to further increase “at risk” knee positions and ACL strain during functional activities common to ACL injury. Examining how these distinct postures influence neuromuscular function of the hip during dynamic activity may help clarify the role of lower extremity posture as a potential injury risk factor for ACL injury.

Single Leg Squat

While injury prevention programs have been developed to target the posterio-lateral hip musculature (Hewett et al., 1999), functional screening tools to appropriately identify those with decreased function of the hip musculature are not readily available. Clinically, the single leg squat is a controlled functional task that is commonly used to assess hip muscle control and dynamic knee alignment, but limited research has quantified the relationship between hip muscle control and lower extremity motion during this task. Although one study observed no relationship between isometric strength of the hip abductors and the amount of hip adduction during the single leg squat (DiMattia, Livengood, Uhl, Mattacola, & Malone, 2005), this study only used 2 dimensional analyses to assess a motion that occurs in 3 planes. In a separate study that examined activation of the posterio-lateral hip musculature during a single leg squat, females demonstrated less gluteus medius and greater gluteus maximus activation compared to males, however this difference was not significant (Zeller et al., 2003). This study was limited to a small sample size (9 males and 9 females) and there was no attempt to control the rate and depth of the squat or the motion at the trunk, which could have directly influenced the activation of the gluteus medius (Schmitz et al., 2002). Studies are therefore needed to examine the validity of predicting hip muscle function during the single leg squat using adequate controls and sample sizes.

Summary

The goal of this review of literature was to provide a rationale for considering the influence of sex differences in lower extremity posture on activation of the hip musculature while providing justification for the use of a single leg squat to identify those that may have decreased activation of the hip musculature during dynamic activity. The gluteus medius and gluteus maximus function as pelvic stabilizers and are essential in control of rotation at the hip and knee, particularly in a single leg stance. Hence, factors that reduce the muscular efficiency and force producing capabilities of the muscles may decrease control of the lower extremity during dynamic activities.

While sex differences in function of the posterio-lateral hip musculature have been suggested to contribute to functional valgus collapse and the increased prevalence of ACL in females, reasons to explain this sex difference are unknown. Sex differences in lower extremity posture characteristics have been observed in the hip and knees, which have the potential to influence the function of the posterio-lateral hip musculature via changes in the moment arms of the muscles. Relationships among these lower extremity posture characteristics and their interactive effects on neuromuscular control provides a strong rationale for the need to examine a comprehensive set of lower extremity posture characteristics as they relate to posterio-lateral hip muscle function and functional valgus collapse. While the single leg squat is thought to be an ideal clinical assessment tool to identify these relationships, more work is needed to quantify this relationship and examine the validity of the single leg squat in predicting hip muscle function.

CHAPTER III

METHODS

The *overall objective* of this research was to determine the effects of static lower extremity posture on hip strength and their collective influence on hip and knee kinematics during a single leg squat. The *central hypothesis* was that a relative valgus static lower extremity posture (characterized by greater standing pelvic angle, hip anteversion, tibiofemoral angle, quadriceps angle and genu recurvatum) would predict decreased hip strength, and collectively, greater relative valgus static lower extremity posture and decreased hip strength would predict greater functional valgus collapse (characterized by increased hip adduction, internal rotation and knee valgus motion) during a single leg squat. A secondary hypothesis is that these relationships would be more pronounced in females compared to males.

The *approach* was to first describe the relationship between static lower extremity posture and hip strength in a cohort of males and females who are known to have differences in pelvic, hip and knee lower extremity posture characteristics. Static lower extremity posture and hip strength was then used to predict lower extremity kinematics during a single leg squat. Structural equation modeling using path analysis determined the extent to which lower extremity posture explained isometric hip torque, and the extent to which lower extremity posture and hip strength collectively explained lower extremity motion during the single leg squat. Separate multivariate analyses of variance

(MANOVAs) compared males and females on lower extremity posture, hip muscle activation and lower extremity motion during the single leg squat. The *rationale* for examining the influence of static lower extremity posture and hip strength on functional valgus collapse in males and females was to identify factors that may allow us to better screen for individuals who may have decreased muscular control of the hip, which has been identified as a potential risk factor for ACL injury. Further, examining the effects of static posture on dynamic control of the hip and knee during a functional task will help us better understand the relationship between static posture and lower extremity motions that are known to strain and injure the ACL.

Subjects

Thirty one healthy males and 31 healthy females, predominantly college students, between the ages of 18 and 35 were recruited from the University and surrounding community to participate in the study. Inclusion criteria for the study were no history of surgery in either lower extremity, no previous hip or knee joint injury within the last 6 months, and no current injury to the lower extremity that would detract from the ability to perform a single leg squat. Subjects read and signed a consent form approved by the University's Institutional Research Board for the protection of human subjects prior to participation. (Appendix A) Subjects also completed an activity rating scale adapted from Marx et al (2001). (Appendix B)

Instrumentation

A Biodex System 3 isokinetic dynamometer (Biodex Medical Systems Inc.; Shirley, NY) was used to record maximal voluntary isometric hip abduction and extension contraction forces. A 16 channel Myopac telemetric system (Run Technologies, Mission Viejo, CA) recorded surface electromyography (sEMG) activity of the gluteus medius (G_{med}) and gluteus maximus (G_{max}) during the maximal voluntary isometric contractions (MVICs) and during the single leg squat. The Myopac unit has an amplification of 1mV/V with a frequency bandwidth of 10 to 1000Hz, a common mode rejection ratio of 90dB min at 60Hz, an input resistance of 1 MΩ, and an internal sampling rate of 8 KHz. The sEMG signal was detected with 10 mm bipolar Ag-AgCl surface electrodes (Blue Sensor N-00-S; Ambu Products, Ølstykke, Denmark; 44.8x22mm diameter; skin contact size 30x22mm) with a center-to-center distance of 20mm. Myoelectric data were acquired, stored and analyzed using DataPac 2K2 lab application software (Version 3.13, Run Technologies, Mission Viejo, CA). Kinematic data for the trunk, pelvis, thigh, shank and foot were sampled at 100 Hz using electromagnetic sensors (Ascension Technology; Burlington, VT) and a Motion Monitor tracking system (Innovative Sports Training; Chicago, IL) during the single leg squat.

Procedures

All testing was completed in one session. After informed consent was obtained, the session began by familiarizing subjects to the procedures for hip abduction and extension strength testing to insure each subject provided a maximal effort during testing.

This was followed by measurement of lower extremity posture (LEP) characteristics. Subjects were then tested for maximum hip abduction and hip extension torques while sEMG recorded maximal voluntary isometric contraction (MVIC) signals of the gluteus medius and maximus for later normalization of the sEMG signal. The session concluded with neuromuscular and kinematic analysis of the hip and knee during a single leg squat. The dominant stance limb was used for all measures as determined by the stance leg used to kick a ball .

Measurement of LEP

After recording age, height and weight, six LEP characteristics were measured on the dominant stance leg of the lower extremity. All standing measures were performed with the feet bi-acromial width apart, toes facing forward, and with subjects looking straight ahead.

Pelvic angle (PA) was measured in standing with the arms crossed over the chest using a modified technique described by Gilliam et al. (1994). The inferior prominence of the anterior superior iliac spine (ASIS) and the most prominent portion of the posterior superior iliac spine (PSIS) were palpated and the angle formed by a line from the ASIS to the PSIS relative to the horizontal plane was measured with an inclinometer (Performance Attainment Associates, St. Paul, Minnesota).

Hip anteversion (HA) was measured using the *Craig's test* (Magee, 1992) in a prone position and the knee flexed to 90°. The hip was passively rotated until the most prominent part of the greater trochanter reached the most lateral position. With the axis of the goniometer positioned on a line between the medial and lateral femoral epicondyles, the angle between the true vertical (verified with a bubble level) and shaft of the tibia was measured using a standard goniometer.

Q-angle (QA) was measured in the frontal plane using a standard goniometer modified with an extension rod attached to the stationary arm to insure accurate alignment with the ASIS. The inferior prominence of the ASIS was palpated, and the subject's finger was carefully and firmly placed over the prominence. The boundaries of the patella and tibial tuberosity were palpated, and the center positions were marked. With the goniometer axis over the patella center, the angle formed by a line from the ASIS to the patella center and a line from the patella center to the tibia tuberosity was measured (Livingston & Mandigo, 1997).

Tibiofemoral angle (TFA) represented the angle formed by the anatomical axes of the femur and tibia in the frontal plane (Moreland et al., 1987). With the goniometer axis over the knee center (midpoint between the medial and lateral joint line in the frontal plane), the stationary arm (modified with an extension rod) was aligned along a line from the knee center to a proximal landmark (defined as the midpoint between the ASIS and the most prominent aspect of the greater trochanter), and the movable arm was aligned along a line from the knee center to a distal landmark (defined as the midpoint between the medial and lateral malleoli).

Genu recurvatum (GR) was measured in supine with a bolster placed under the distal tibia. The subject was instructed to actively contract the quadriceps to maximally extend the knee. A standard goniometer with the axis over the lateral femoral epicondyle was used to measure the angle between the stationary arm and movable arm aligned with the greater trochanter and lateral malleolus respectively.

Navicular drop (ND) was measured in standing using a modified technique described by Brody (1982). Using a straight edge ruler, the difference between the heights of the navicular tubercle measured in subtalar joint neutral and in relaxed stance determined the amount of navicular drop. Subtalar joint neutral was identified as the position where the medial and lateral aspects of the talar head were equally palpable as subjects inverted and everted the hindfoot. All measurements were taken 3 times, and recorded to the nearest degree or millimeter. Table 1 illustrates results from previous work confirming the investigator's good to excellent day to day reliability on all measures with ICCs $\geq$ .87 (Shultz, Nguyen et al., 2006).

Table 1. Reliability of LEP Measurements

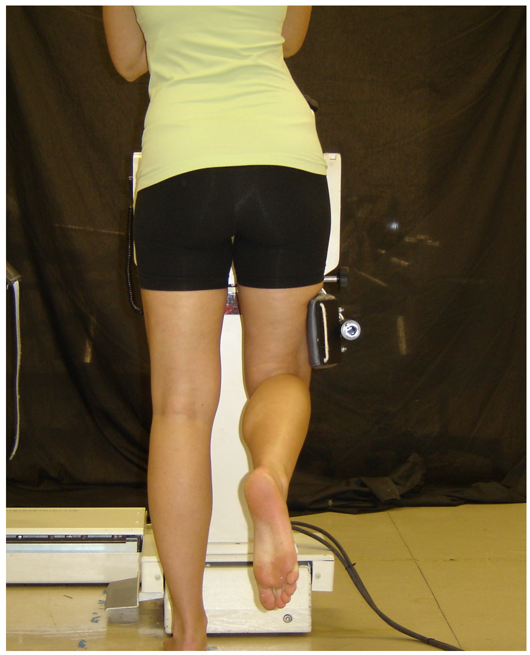
Anatomical Measure	**$ICC_{2,k}$ (SEM)**
Pelvic Angle (deg)	.98 (0.5)
Hip Anteversion (deg)	.97 (1.1)
Standing Q-Angle (deg)	.98 (0.8)
Tibiofemoral Angle (deg)	.87 (0.7)
Genu Recurvatum (deg)	.97 (0.5)
Navicular Drop (mm)	.97(0.4)

Measurement of Hip Torque and MVIC Myoelectrical Signals

sEMG electrodes were placed over the G_{med} and the G_{max} of the dominant limb according to procedures described by Cram and Kasman (1998). Electrodes were placed on the G_{med} at a position one third the distance from the greater trochanter to the iliac crest. Electrode placement of the G_{max} was positioned midway between the greater trochanter and the first sacral vertebrae. sEMG electrodes were oriented perpendicular to the length of the muscle fibers and placed over the mid-belly. The reference electrode was secured to the medial aspect of the tibia. Prior to attaching the electrodes, all skin areas were thoroughly cleaned with isopropyl alcohol.

A modification of a technique described by Carcia et al. (2005) was used to assess functional hip abduction torque and to acquire the MVIC myoelectric signal of the G_{med}. (Figure 3) Subjects stood adjacent to the Biodex dynamometer with the trunk erect, feet facing forward, arms crossed over the chest, and looking straight ahead. The dynamometer axis was aligned with the head of the femur determined by the intersection of a medially directed horizontal line from the greater trochanter and a distally directed vertical line from the anterior superior iliac spine (Nyland, Smith, Beickman, Armsey, & Caborn, 2002). The resistance arm of the dynamometer was positioned on the lateral side of the non-stance leg with the distal edge of the pad approximately 5 cm proximal to the lateral joint line and the hip positioned in approximately 5° of abduction. Subjects performed 3 trials of 3 second MVICs by abducting the hip while supporting their body weight on the dominant stance limb and maintaining an erect posture. A 30 second rest period separated each trial with subjects resting in a bilateral stance. The decision for performing functional hip abduction while standing on the dominant stance limb was based on a pilot study comparing activation between the stance and non-stance limb in 10 healthy subjects. Paired samples t-test used to examine these pilot data demonstrated significantly higher peak sEMG amplitudes of the G_{med} in the stance limb compared to the non-stance limb (.369 vs. .275 volts, p>.001).

Figure 3. Subject Position for Standing MVIC Testing of Hip Abduction



To assess hip extension torque and acquire MVIC myoelectric signals of the G_{max}, subjects performed extension of the hip in a supine position with the hip flexed to 90º and the dynamometer axis aligned with the greater trochanter. The resistance arm was positioned on the posterior thigh just proximal to the knee joint line. (Figure 4) Since there was no accepted method to assess MVIC of hip extension, this procedure was determined through extensive pilot testing. Subjects performed 3 trials of 3 second hip extension MVICs. A 30 second rest period separated each trial. Results of pilot work to assess day to day reliability of sEMG amplitude and isometric torque production for these muscles using the previously described protocols are summarized in Table 2, and demonstrate good to excellent measurement consistency across days. (ICC$\geq$ .83).

Figure 4. Subject Position for MVIC Testing of Hip Extension

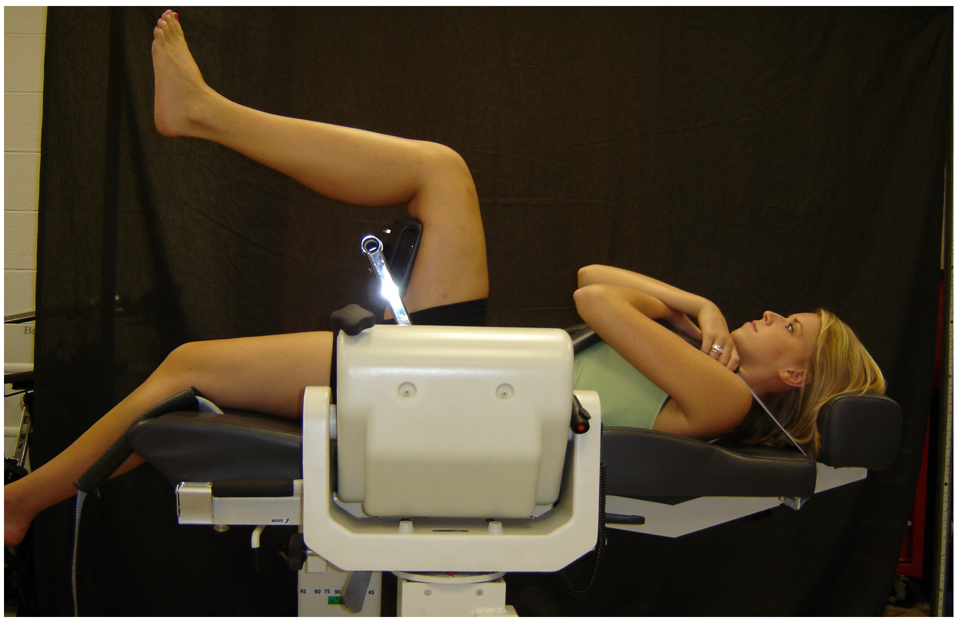


Table 2. Previously Established Reliability of MVIC Myoelectrical Signals and Hip Torque

Measure	**ICC$_{2,k}$ (SEM)**
sEMG G_{med} with Abduction *(volts)*	.97 (.04)
sEMG G_{max} with Extension *(volts)*	.97 (.09)
Abduction Torque *(Nm/kg)*	.91 (.03)
ExtensionTorque *(Nm/kg)*	.89 (.46)

Measurement of Joint Motion during the Single Leg Squat

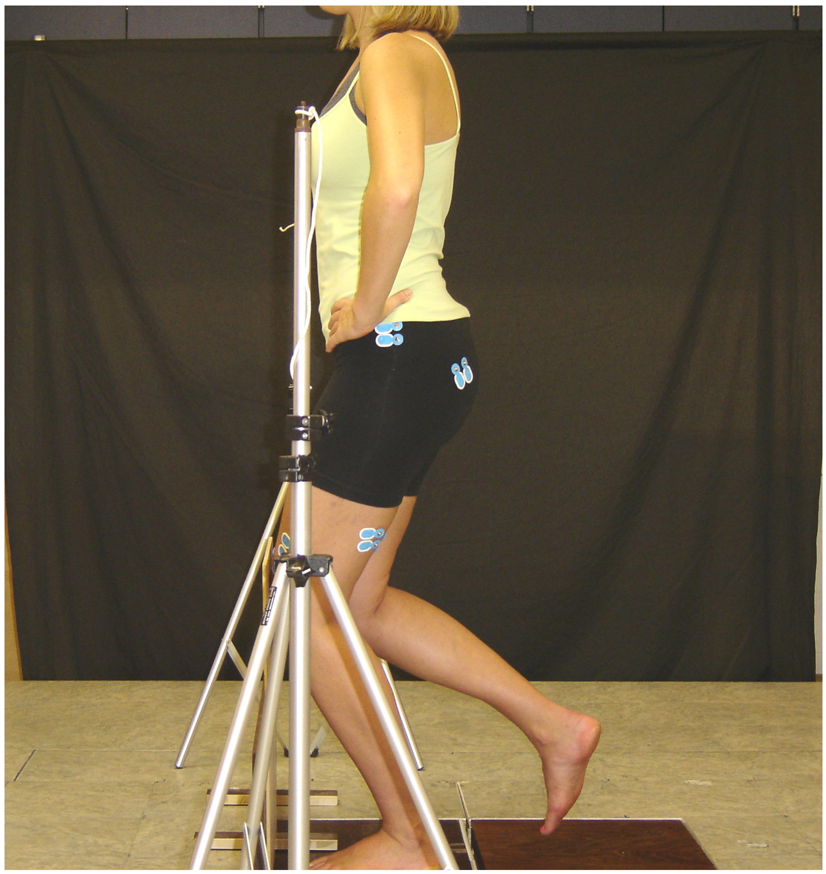
Subject set up: With sEMG electrodes still attached, motion sensors were secured to the foot, tibial shaft, the lateral thigh, sacrum and thorax to obtain positions and orientation of each rigid segment. Digitization procedures were performed using the default selection with a segmental reference system defining body segments with the positive X-axis defined as the posterior to anterior axis; positive Y-axis defined as the distal to proximal longitudinal axis; and positive Z-axis defined as the medial to lateral axis. The ankle joint center was determined by the midpoint between the medial and lateral malleoli, the knee joint center by the midpoint between the medial and lateral joint line, and the hip joint center was determined by the Leardini method (Leardini et al., 1999). Hip and knee angles were calculated using Euler angle definitions with a rotational sequence of Z X' Y".

Single leg squat: Subjects stood in a starting position with feet shoulder width apart, hips and knees extended, toes facing forward, equal weight on both feet and thumbs lightly touching the iliac crests. A plywood board was positioned at a distance anterior to the knee while subjects performed a double leg squat to 60° of knee flexion based on real time goniometer values. The plywood board was positioned to provide subjects feedback indicating they had reached 60° of knee flexion during each trial. Subjects then performed a single leg squat with instructions to squat straight down until they touched the board with their knee while looking straight ahead, and to maintain an upright position without flexing the trunk forward or to the side. A string was positioned perpendicular to the 1st toes at the level of the chest to monitor forward flexion of the trunk. (Figure 5) Subjects were instructed to keep an upright posture in order to limit the influence of trunk motion on the hip musculature, particularly the G_{med} (Schmitz et al., 2002).

Figure 5. Set-up for Single Leg Squat

Each single leg squat trial was initiated by a verbal command from the examiner and performed at a speed of five seconds from the starting position to 60° of knee flexion. The rate of the task was controlled by a metronome set at a cadence of 60 beats per minute. Subjects transitioned from bilateral stance to single leg stance during the first two beats, with the non-stance knee and hip flexed approximately 45° and 0° respectively. The squat then began on the third beat and ended at 60° of knee flexion on the fifth beat (total squat time 2 seconds). (Figure 6)

Figure 6. Sagittal View of Single Leg Squat with Representative sEMG Placement



A force plate marked the transition from double leg stance to single leg stance (start of trial) and 60° of knee flexion marked the end of the trial. Subjects were allowed sufficient practice to ensure the task was performed properly, and data were then collected during 5 acceptable trials. A trial was deemed unacceptable if subjects: 1) touched the string (indicating increased forward flexion of the trunk), 2) touched the non-stance leg to the ground or stance leg, 3) lifted either hand off the iliac crest, or 4) failed to reach 60º of knee flexion as confirmed by real time goniometry. Pilot testing of 10 healthy females (24.7 ± 2.3 yrs, 173.4 ± 11.6 cm, 66.1 ± 9.1 kg) using these procedures revealed excellent day to day reliability for the lower extremity joint motions and sEMG activity (ICC >.86), indicating subjects could consistently perform the task (Table 3). A consistent mean forward trunk flexion angle of 7.6°±3.4 confirmed subjects could maintain an upright trunk during the task.

Table 3. Previously Established Reliability of MVIC Myoelectrical Signals and Kinematic Measures During a Single Leg Squat

Measure	**$ICC_{2,k}$ (SEM)**
G_{med} sEMG Activation *(volts)*	.94 (.04)
G_{max} sEMG Activation *(volts)*	.96 (.01)
Hip Adduction Excursion *(degrees)*	.93 (2.5)
Hip Internal Excursion *(degrees)*	.86 (1.9)
Knee Valgus Excursion *(degrees)*	.91 (2.8)
Knee Flexion Excursion *(degrees)*	.97 (1.9)
Knee Rotation Excursion *(degrees)*	.92 (1.4)
Trunk Flexion Excursion *(degrees)*	.90 (1.3)

Data Reduction

The three measures of each LEP characteristic were entered into an Excel spreadsheet and the average of the three measurements for each variable was calculated and used for analyses. Biodex torque data were recorded as the maximum peak torque obtained from 3 MVIC trials each for hip abduction and extension, and entered into an Excel spreadsheet. Peak toque was then normalized to the subject's body mass and reported in Newton-meters per kilogram of body mass (Nm/kg). Kinematic hip and knee angles in the coronal, transverse, and sagittal planes were exported from the Motion Monitor Software into an Excel spreadsheet. Initial joint angles were calculated as the average joint positions during the first second following transition from double leg to single leg stance. Final joint angles were determined as the value when subjects achieved 60° of knee flexion. Joint excursion was calculated as the difference (final minus initial) for each trial and the average across 5 trials was used for statistical analysis.

sEMG of the G_{med} and G_{max} during the MVIC and single leg squat trials were filtered from 10 Hz to 350 Hz, using a fourth-order, zero-lag Butterworth filter then processed using a centered root mean square algorithm with 100 ms time constant. The peak RMS value obtained over 3 MVIC trials for each muscle was used to normalize the sEMG data during the single leg squat. The average RMS amplitude of the 5 single leg squat trials for the first 20% of the trial (single leg stance), the last 20% of the trial (prior to 60º knee flexion), and across the entire trial (after transition to single leg weight bearing to 60º) was then normalized to the subject's MVIC peak RMS value and reported as a percentage of the MVIC (%MVIC).

Statistical Analyses

1. To test hypothesis 1a and 1b, path analysis, as implemented by structural equation modeling was used to examine 1) whether greater valgus and pronation postures predicted decreased hip abduction torque and 2) whether greater valgus posture and decreased hip abduction torque collectively predicted greater dynamic valgus knee motion (increased hip adduction and internal rotation, and knee external rotation and valgus) as measured kinematically during the single leg squat, once accounting for G_{med} activation.
2. To test hypothesis 2a and 2b, a separate path analysis, as implemented by structural equation modeling was used to examine 1) whether greater valgus and pronation postures predicted decreased hip extension torque and 2) whether greater valgus posture and decreased hip extension torque collectively predicted greater dynamic valgus knee motion (increased hip adduction and internal rotation, and knee external rotation and valgus) as measured kinematically during a single leg squat, once accounting for G_{max} activation.
3. To test hypothesis 3, separate multivariate analyses of variance (MANOVAs) determined whether females and males differed on measures of LEP (2 [sex] x 2 [posture]), hip strength (2 [sex] x 2 [hip torque]), and total hip and knee motion during the single leg squat (2 [sex] x 4 [joint excursion]).

CHAPTER IV

RESULTS

Sixty-two subjects successfully completed data collection. However, data on two subjects were not appropriately recorded due to unexpected computer problems. Therefore, data from 30 males and 30 females (males: age=23.9$\pm$3.6 yrs, height=178.5$\pm$9.9 cm, mass=82.0$\pm$14.1 kg; females: age=22.2$\pm$2.6 yrs, height=162.4$\pm$6.3 cm, mass=60.3$\pm$8.1 kg) were used for analyses. Mean$\pm$SD, median and range (minimum to maximum) for measures of lower extremity alignment and hip torque on the dominant stance limb are listed in Table 4 while descriptive statistics for joint excursion and posterio-lateral hip activation during the SLS are listed in Table 5. Histograms for all predictor and dependent variables are presented in Appendices C_1-C_{12}, which appear to demonstrate relatively normal distributions and reasonable ranges for each variable. Measurements were taken on the left limb in fifty-five subjects and the right limb in five subjects. The complete set of raw data for subject demographics, predictor variables and dependent variables can be found in Appendices E-G.

Table 4. Descriptive Statistics for Activity Rating Scale, Lower Extremity Posture and Hip Torque

Measure	Mean (SD)	Median	Range
Activity Rating Scale	8.4 (4.1)	8.5	0-16
Pelvic Angle *(degrees)*	11.1 (4.6)	11	0.0-21.0
Hip Anteversion *(degrees)*	10.7 (5.2)	9.8	1.0-27.7
Quadriceps Angle *(degrees)*	12.9 (5.6)	12	1.0-29.0
Tibiofemoral Angle *(degrees)*	10.7 (2.0)	10.7	5.0-15.3
Genu Recurvatum *(degrees)*	3.8 (3.8)	3.0	-1.3-14.3
Navicular Drop *(millimeters)*	6.6 (6.0)	6.3	-4.0-25.7
Hip ABD Torque *(Nm/kg)*	0.69 (.19)	0.66	0.37-1.33
Hip EXT Torque *(Nm/kg)*	3.46 (1.05)	3.43	1.87-5.80

Table 5. Descriptive Statistics for Joint Excursion and sEMG

Measure	Mean (SD)	Median	Range
Hip Adduction *(degrees)*	11.4 (10.4)	12.0	-15.3-35.5
Hip Internal Rotation *(degrees)*	-2.3 (5.9)	-1.6	-16.4-12.8
Knee Valgus *(degrees)*	-0.1 (8.0)	-0.4	-23.5-17.0
Knee External Rotation *(degrees)*	2.7 (6.1)	2.2	-9.8-20.2
G_{med} sEMG Initial 20% *(% MVIC)*	0.20 (0.10)	0.17	0.06-0.58
G_{med} sEMG Final 20% *(% MVIC)*	0.33 (0.15)	0.30	0.11-0.80
G_{med} sEMG Total *(% MVIC)*	0.27 (0.13)	0.23	0.11-0.72
G_{max} sEMG Initial 20% *(% MVIC)*	0.19 (0.22)	0.12	0.02-1.14
G_{max} sEMG Final 20% *(% MVIC)*	0.21 (0.19)	0.16	0.03-1.00
G_{max} sEMG Total *(%MVIC)*	0.20 (0.19)	0.14	0.03-1.04

As the valgus postural factor was based on data from previous work (Nguyen & Shultz, In Review), preliminary factor analysis was first performed to examine the validity of the valgus posture factor for this cohort, and revealed insufficient relationships among the alignment variables. Based on this analysis, it was determined that co-variation among the variables in the current subset of subjects may be insufficient to identify the postural relationships, and therefore would be inappropriate to collapse the variables into a single valgus postural factor. Therefore, the lower extremity alignment variables of pelvic angle, hip anteversion, quadriceps angle, tibiofemoral angle and genu recurvatum were entered into the model as separate observed variables, not as factor loadings on the latent variable of valgus posture.

Preliminary analyses also revealed high correlations between the normalized RMS amplitudes for the first 20% of the trial (single leg stance), the last 20% of the trial (prior to 60º knee flexion), and across the entire trial for both the G_{med} ($r = .80$-$.96$) and G_{max} ($r = .91$-$.98$). Therefore, consistent with the method used to calculate joint excursions, the normalized RMS amplitude (% MVIC) of the posterio-lateral hip musculature across the entire trial (after transition to single leg weight bearing to 60º of knee flexion) was used for data analysis.

<u>Relationships Between LEP, Hip Abduction Torque, and Functional Valgus Collapse</u>

A path analysis was used to examine 1) the extent to which static lower extremity alignment (PA, HA, QA, TFA, GR, ND) predicted hip abduction torque and 2) to examine their collective influence on dynamic alignment (hip adduction, hip internal rotation, knee valgus, knee external rotation) during a single leg squat. A full path model examining the correlation matrix among twelve variables (Table 6), six predictor variables (LEA) and six dependent variables (abduction strength and dynamic valgus while accounting for G_{med} activation), was performed with all direct and indirect paths specified. (Figure 7) Inferential goodness-of-fit index indicated the full model was a perfect fit ($\chi^2 = 0.00$, df = 0, p = 1.00, RMSEA =0.00), due to the model being saturated with zero degrees of freedom. However, the parameter estimates of the full path model examining all 12 variables was highly unreliable as the total number of parameters being estimated was greater than the total sample size.

To examine a more stable model, the full model was modified by first removing the dependent measures that had no statistically significant paths (i.e. variables that had no significant predictors). Statistical significance was determined by the *t*-value statistic which reflects the ratio of the parameter estimate to its standard error. A *t*-value greater than +2 or less than -2 is considered statistically significant (Raykov & Marcoulides, 2000). The dependent measures that were removed included hip adduction, hip internal rotation and knee valgus. The resultant model was further modified by removing the predictor variables that did not approach significance or were non-significant in explaining any of the remaining outcome measures (dependent variables). These included measures of pronation and pelvic angle. The final modified model is illustrated in Figure 8.

Table 6. Correlation Matrix of Predictor and Dependent Variables for Hypotheses 1a and 1b

	1	2	3	4	5	6	7	8	9	10	11	12
1. G_{med} Activation	1											
2. Hip ADD	.127	1										
3. Hip IR	-.001	-.260*	1									
4. Knee VAL	-.004	.582*	-.555*	1								
5. Knee ER	.243	.503*	-.120	.492*	1							
6. Abduction Torque	-.275*	-.066	-.091	-.077	-.369*	1						
7. Navicular Drop	.086	.021	.253	.088	.002	.063	1					
8. Pelvic Angle	.019	-.051	-.176	.233	-.170	.076	.261*	1				
9. Hip Anteversion	.006	-.077	.097	.089	.244	-.263*	-.171	.162	1			
10. Quadriceps Angle	-.048	-.121	.042	.125	.018	-.148	.067	.467*	.234	1		
11. Tibiofemoral Angle	.096	.123	-.143	.134	.200	-.067	-.207	.064	-.089	.171	1	
12. Genu Recurvatum	.170	.115	.142	-.004	.204	.056	.406*	-.223	-.175	-.187	-.057	1

N = 60, * significant correlations ($p < 0.05$)

Figure 7. Full Path Model for Dependent Variables Abduction Torque and Functional Valgus Collapse

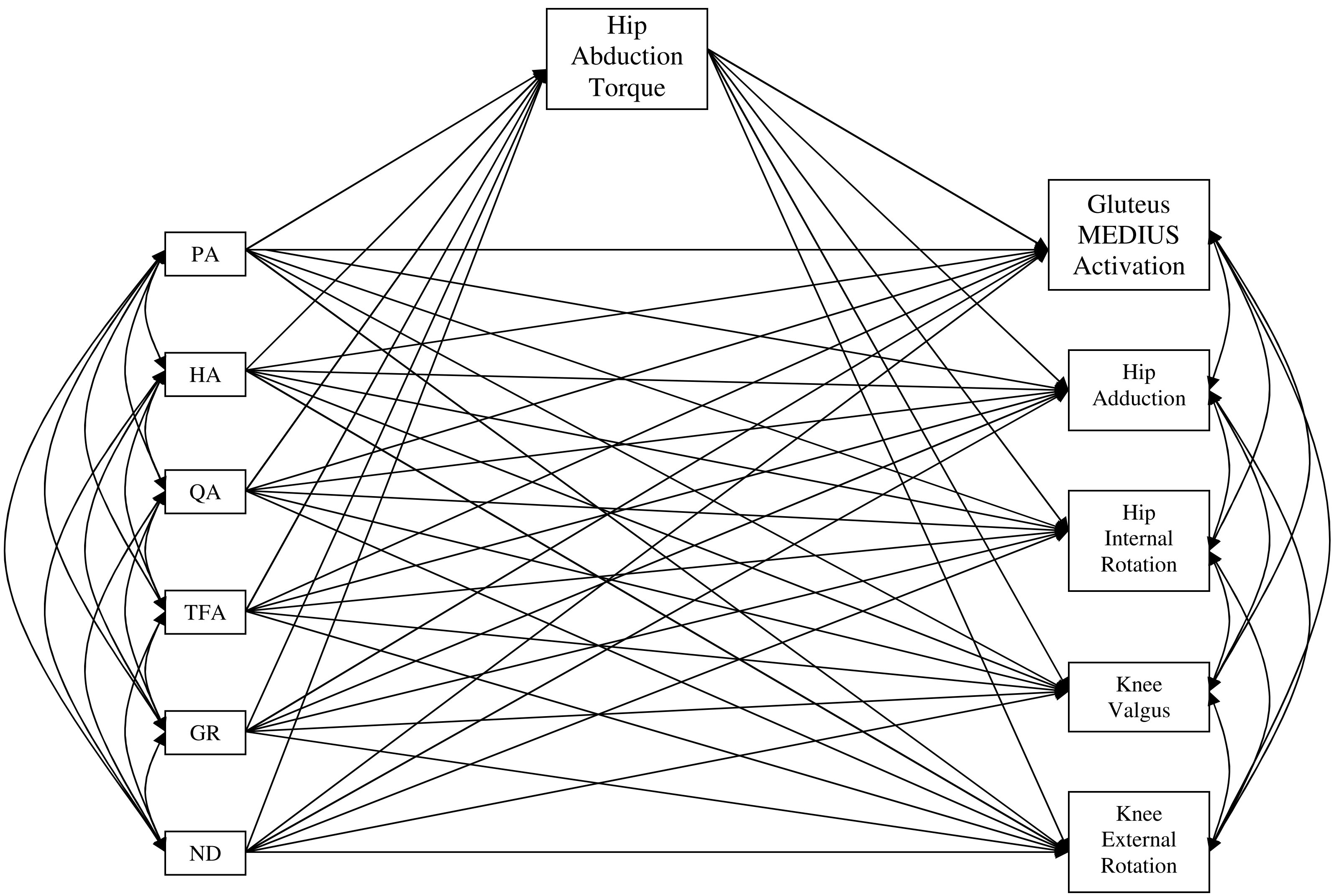
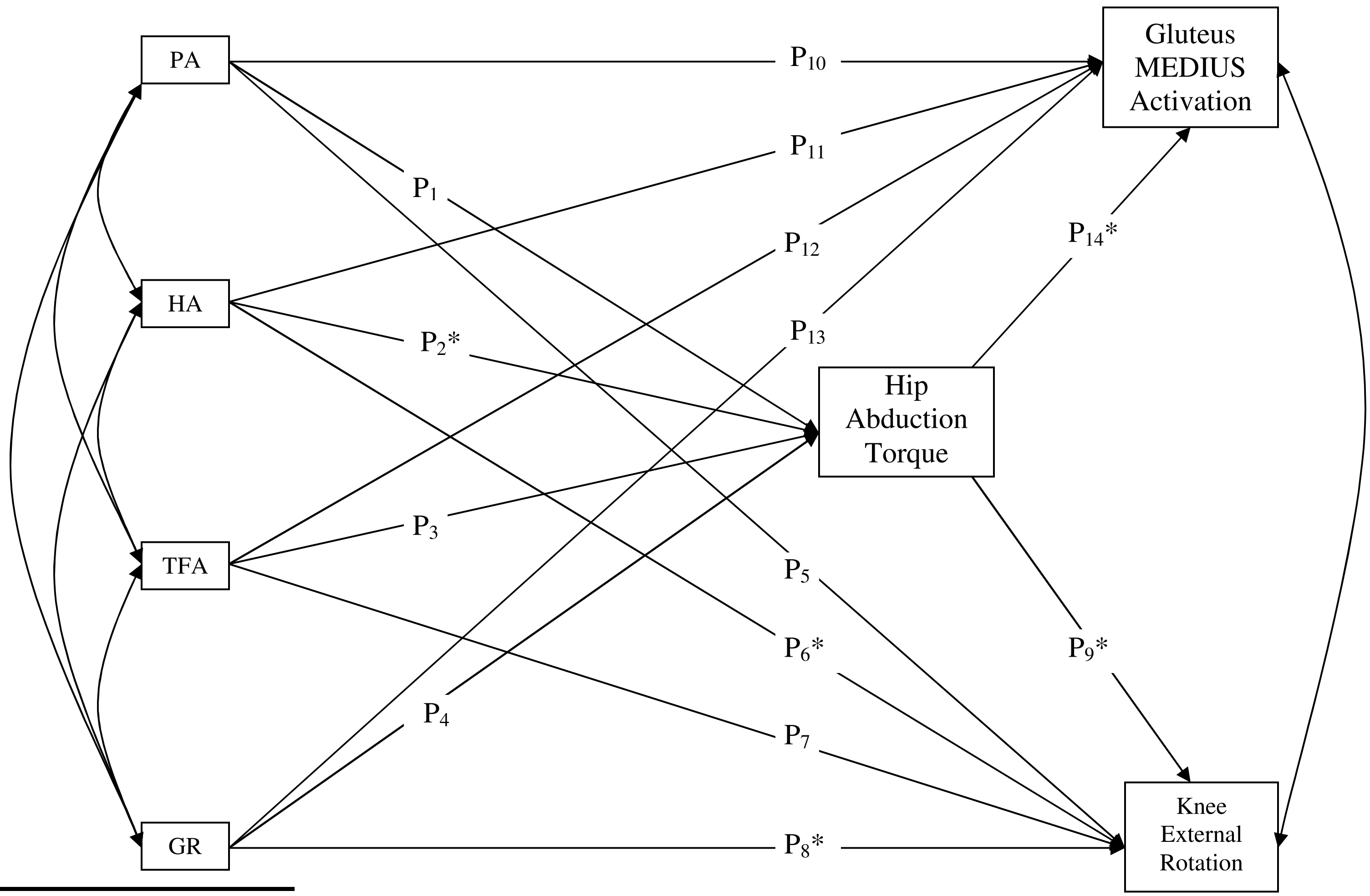


Figure 8. Final Model for Dependent Variables Abduction Torque, Dynamic Valgus, G_{med} Activation



* significant path coefficient

Hypothesis 1a: LEP Predicting Hip Abduction Torque

Table 7 lists the coefficients, standard errors of the coefficients, and t statistics for paths P_1-P_4 that represents the extent to which lower extremity posture (PA, HA, TFA, GR) predicted hip abduction torque. These findings reveal that only 9% of the variance in hip abduction torque can be explained by the model. Hip anteversion (P_2) was the only path coefficient ($\beta = -0.29$) that was statistically significant ($t = -2.18$, $p < .05$) indicating that greater hip anteversion resulted in decreased hip abduction torque, once accounting for pelvic angle, tibiofemoral angle and genu recurvatum.

Table 7. Path Coefficients of LEP Predicting Hip Abduction Torque

	Hip Abduction Torque			
LEP	**Path Coefficient**	**Standard Error**	**t-value**	**p-value**
Pelvic Angle	0.14	0.13	1.02	> .05
Hip Anteversion	-0.29	0.13	-2.18	< .05*
Tibiofemoral Angle	-0.10	0.13	-0.77	> .05
Genu Recurvatum	0.03	0.13	0.23	>.05

* significant path coefficient

Hypothesis 1b: LEP and Hip Abduction Torque Predicting Functional Valgus Collapse

Table 8 lists the coefficients, standard errors of the coefficients, and t statistics for paths P_5-P_9 that represents the extent to which LEP characteristics and hip abduction torque predicted knee external rotation excursion during the single leg squat, once G_{med} activation was accounted for. These findings reveal that 29% of the variance in knee external rotation excursion during the single leg squat can be explained by the model.

The path coefficients for hip anteversion (P_6, t = 2.09, p<.05), genu recurvatum (P_8, t = 2.07, p<.05) and hip abduction torque (P_9, t = -2.42, p<.05) were significant predictors of knee external rotation, with tibiofemoral angle closely approaching significance (P_7, t = 1.97). These results indicate that greater knee external rotation during a single leg squat was predicted by greater hip anteversion and genu recurvatum, and decreased hip abduction torque, once G_{med} activation and other LEP characteristics (pelvic angle and tibiofemoral angle) were accounted for.

Table 8. Path Coefficients of LEP and Hip Abduction Torque Predicting Knee External Rotation

	Knee External Rotation			
	Path Coefficient	**Standard Error**	**t-value**	**p-value**
Pelvic Angle	-0.15	0.12	-1.26	>.05
Hip Anteversion	0.26	0.12	2.09	<.05*
Tibiofemoral Angle	0.23	0.12	1.97	>.05
Genu Recurvatum	0.24	0.12	2.07	<.05*
Hip Abduction Torque	-0.29	0.12	-2.42	<.05*

Summary of Results Specific to Hypotheses 1a and 1b

Modification of the full path analysis model resulted in significant relationships between the predictor variables: hip anteversion, genu recurvatum, and dependent variables: hip abduction torque and knee external rotation. Once the LEP characteristics and G_{med} activation were accounted for, greater hip anteversion predicted decreased hip abduction torque (P_2) while greater hip anteversion (P_6) and genu recurvatum (P_8) and decreased hip abduction torque (P_9) predicted greater knee external rotation during the single leg squat.

The only results from this path analysis model that supports an indirect ("sequential") effect was greater hip anteversion predicting decreased hip abduction torque, which in turn predicted greater knee external rotation during the single leg squat (P_2 and P_9). However, hip anteversion had both a direct effect and indirect effect on knee external rotation. Comparing the direct path of hip anteversion (P_6) and the indirect paths of hip anteversion through hip abduction torque (P_2 and P_9), the direct path coefficient is greater than the product of the indirect paths (0.26 vs 0.08). This suggests that greater static hip anteversion had a stronger direct effect on predicting greater knee external rotation compared to its indirect effect through decreased hip abduction torque.

Relationships Between LEP, Hip Extension Torque, and Functional Valgus Collapse

A separate path analysis was used to examine the extent to which static lower extremity alignment predicted hip extension torque, and then examine their collective influence on dynamic alignment during a SLS while accounting for G_{max} activation. A full path model examining the correlation matrix among twelve variables (Table 9), with hip extension torque and dynamic valgus as the dependent variables (controlling for G_{max} activation) was performed with all direct and indirect paths specified. Inferential goodness-of-fit index indicated the full model was also a perfect fit ($\chi^2 = 0.00$, df = 0, p = 1.00, RMSEA =0.00), due to the model being saturated with zero degrees of freedom.

This full path model was also highly unreliable as the total number of parameters being estimated was greater than the total sample size.

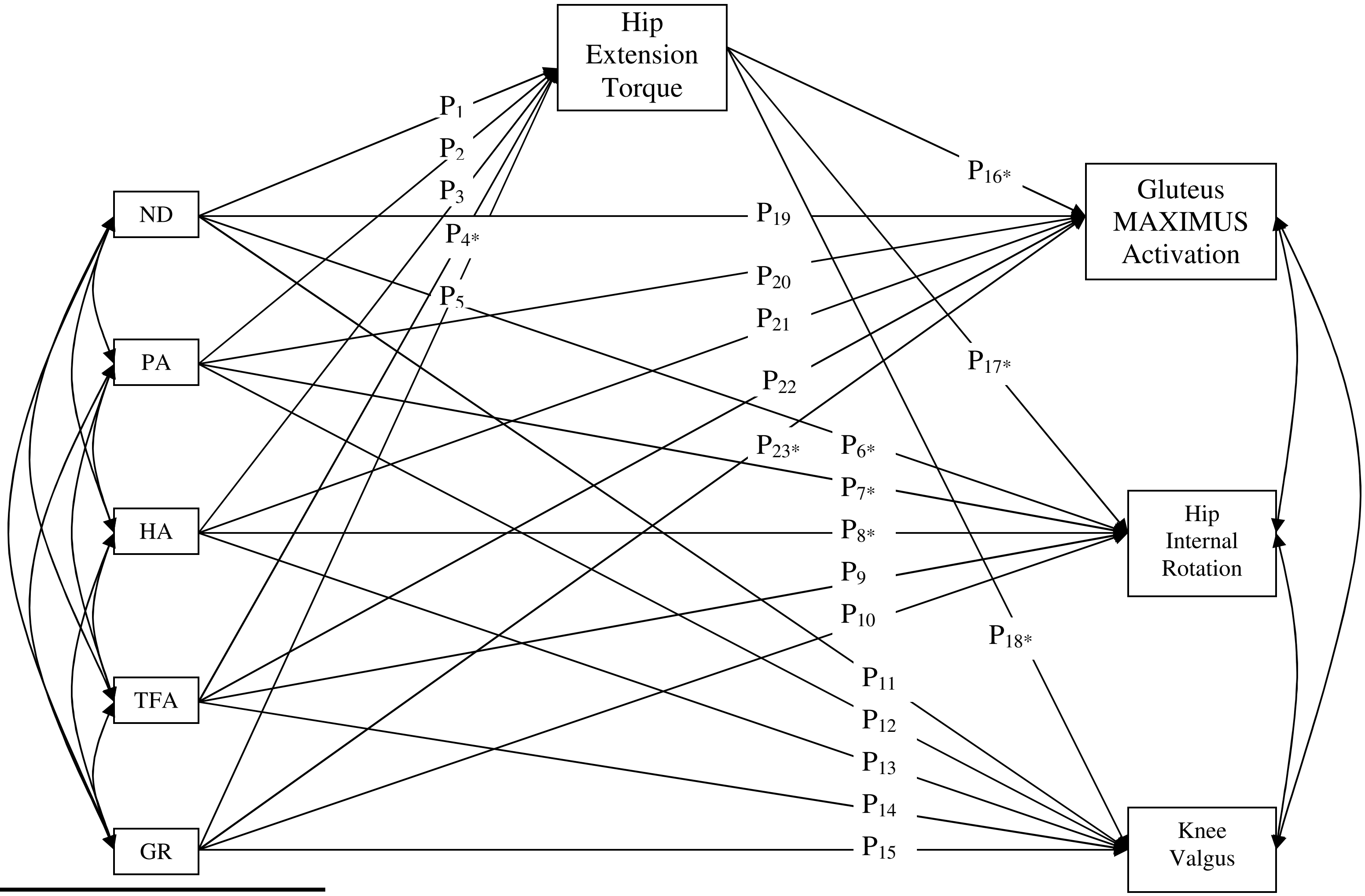
To examine a more stable model, the full model was modified in the same manner as previous described for Hypothesis 1 by first removing the dependent measures that had no statistically significant paths. This resulted in hip adduction and knee external rotation being removed from the model. Further modification of the path analyses included the removal of quadriceps angle as a predictor variable, as it failed to explain any of the remaining outcome measures. The final modified model is illustrated in Figure 9.

Table 9. Correlation Matrix of Predictor and Dependent Variables for Hypotheses 2a and 2b

	1	2	3	4	5	6	7	8	9	10	11	12
1. G_{max} Activation	1											
2. Hip ADD	.100	1										
3. Hip IR	-.320*	-.260*	1									
4. Knee VAL	.340*	.582*	-.555*	1								
5. Knee ER	.020	.503*	-.120	.492*	1							
6. Extension Torque	-.612*	-.087	.242	-.382*	-.235	1						
7. Navicular Drop	-.032	.021	.253	.088	.002	-.102	1					
8. Pelvic Angle	.169	-.051	-.176	.233	-.170	-.145	.261*	1				
9. Hip Anteversion	.219	-.077	.097	.089	.244	-.179	-.171	.162	1			
10. Quadriceps Angle	.069	-.121	.042	.125	.018	-.194	.067	.467*	.234	1		
11. Tibiofemoral Angle	.040	.123	-.143	.134	.200	-.307*	-.207	.064	-.089	.171	1	
12. Genu Recurvatum	-.176	.115	.142	-.004	.204	-.155	.406*	-.223	-.175	-.187	-.057	1

N = 60, * significant correlations ($p < 0.05$)

Figure 9. Final Model for Dependent Variables Extension Torque, Dynamic Valgus, G_{max} Activation



* significant path coefficient

Hypothesis 2a: LEP Predicting Hip Extension Torque

Table 10 lists the coefficients, standard errors of the coefficients, and t statistics for paths P_1-P_5 that represents the extent to which LEP characteristics (ND, PA, HA, TFA, GR) predicted hip extension torque. These findings reveal that 21% of the variance in hip extension torque can be explained by the model. While the effect of hip anteversion (path P_3) closely approached significance ($t = 1.95$), tibiofemoral angle (path P_4) was the only path that was statistically significant ($t = -2.83$, $p < .05$) indicating that greater tibiofemoral angle predicted decreased hip extension torque once all other LEP characteristics were accounted for.

Table 10. Path Coefficients of LEP Predicting Hip Extension Torque

LEA	Path Coefficient	Standard Error	t-value	p-value
Navicular Drop	-0.12	0.15	-0.76	> .05
Pelvic Angle	-0.09	0.14	-0.68	> .05
Hip Anteversion	-0.25	0.13	-1.95	>.05
Tibiofemoral Angle	-0.36	0.13	-2.83	< .05*
Genu Recurvatum	-0.19	0.14	-1.35	>.05

* significant path coefficient

Hypothesis 2b: LEP and Hip Extension Torque Predicting Functional Valgus Collapse

Table 11 lists the coefficients, standard errors of the coefficients, and t statistics for paths P_6-P_{18} that represents the extent to which LEP characteristics and hip extension torque predict hip internal rotation and knee valgus excursion during a single leg squat while accounting for G_{max} activation. These findings reveal that 25% of the variance in hip internal rotation and 18% of the variance in knee valgus during the single leg squat can be explained by the model. Once accounting for other LEP characteristics (tibofemoral angle and genu recurvatum) and G_{max} activation, path coefficients for navicular drop (P_6, t = 2.86, p<.05), pelvic angle (P_7, t = -2.12, p<.05), hip anteversion (P_8, t = 2.21, p<.05) and hip extension torque (P_{17}, t = 2.44, p<.05) were significant in predicting hip internal rotation. The only significant path that predicted knee valgus was hip extension torque (P_{18}, t = -2.56, p<.05). These results indicate that greater hip internal rotation during the single leg squat was predicted by greater navicular drop, hip anteversion, hip extension torque and decreased pelvic angle. Decreased hip extension torque was the only predictor of increased knee valgus during the single leg squat. There were no LEP variables that directly predicted dynamic knee valgus during the single leg squat.

Table 11. Path Coefficients of Hip Extension Torque Predicting Hip Internal Rotation and Knee Valgus

	Hip Internal Rotation				Knee Valgus			
	Path Coeff.	Stand. Error	t value	p value	Path Coeff.	Stand. Error	t value	p value
Navicular Drop	0.42	0.15	2.86	<.05*	0.02	0.16	0.15	>.05
Pelvic Angle	-0.29	0.14	-2.12	<.05*	0.17	0.14	1.17	>.05
Hip Anteversion	0.28	0.13	2.21	<.05*	0.00	0.13	-0.01	>.05
Tibiofemoral Angle	0.09	0.13	0.67	>.05	0.02	0.14	0.13	>.05
Genu Recurvatum	0.01	0.14	0.07	>.05	-0.03	0.15	-0.21	>.05
Hip Extension Torque	0.32	0.13	2.44	<.05*	-0.35	0.14	-2.56	<.05*

* significant path coefficient

Summary of Results Specific to Hypotheses 2a and 2b

Modification of the full path analysis model resulted in significant relationships between the predictor variables: navicular drop, pelvic angle, hip anteversion, tibiofemoral angle, genu recurvatum, and dependent variables: hip extensor torque, hip internal rotation and knee valgus. Once the LEP characteristics and G_{max} activation were accounted for, greater tibiofemoral angle predicted decreased hip extension torque (P_4). Greater hip internal rotation during the single leg squat was predicted by greater navicular drop (P_6), greater hip anteversion (P_7), greater hip extension torque (path P_{17}) and decreased pelvic angle (P_8). Decreased hip extension torque was the only predictor of greater knee valgus (P_{18}) during the single leg squat.

The only results from this path analysis model that supports an indirect ("sequential") effect was greater tibiofemoral angle predicting decreased hip extension torque, which in turn predicted greater knee valgus during the single leg squat (P_4 and P_{18}). While the significant paths of greater navicular drop and hip anteversion directly predicted greater hip internal rotation during a single leg squat (P_6 and P_8), they did not act indirectly through decreasing hip extension torque. Interestingly, greater hip internal rotation during the single leg squat was predicted by decreased pelvic angle and greater hip extension torque which is opposite of the relationship hypothesized.

Hypothesis 3: Sex Differences in LEP, Hip Strength and Functional Valgus Collapse

Mean (SD), median and range (minimum to maximum) statistics for all variables by sex are summarized in Table 12. Separate MANOVAs were used to examine sex differences in LEP variables, hip strength, and total hip and knee motion during the single leg squat. Multivariate tests revealed a significant effect of sex with the test statistics performed for each of the MANOVAs (all $p<.001$). Table 13 summarizes the effect of sex on each dependent variable and their respective effect size. Effect size (d) for each variable was calculated as the mean difference between sexes divided by the pooled standard deviation (SD_p). Since the sample sizes were equal for males and females, the pooled standard deviation was calculated as the sum of the standard deviation for males and females divided by 2. The following equations were used to calculate effect size:

$$SD_p = \frac{SD_{males} + SD_{females}}{2} \qquad d = \frac{Mean_{males} - Mean_{females}}{SD_p}$$

Results examining sex differences in LEP characteristics indicate that females had greater mean hip anteversion, quadriceps angle and tibiofemoral angle. Females were also found to have less normalized hip extension torque than males. The only joint motion during the single leg squat that was different between sexes occurred at the knee, where females moved into greater knee valgus compared to males. No sex differences were observed in mean values of other LEP characteristics (pelvic angle, genu recurvatum, navicular drop), hip abduction torque, or joint motions of hip adduction, hip internal rotation and knee external rotation during the single leg squat. However, moderate effect sizes (Howell, 2002) were noted for pelvic angle (0.47), genu recurvatum (0.48), hip abduction torque (0.41) and knee external rotation excursion (0.52), all of which neared significance ($p = .052 - .121$).

Table 12. Descriptive Statistics for Lower Extremity Alignment, Hip Torque and Joint Excursion by Sex

	Males			Females		
Measure	**Mean (SD)**	**Median**	**Range**	**Mean (SD)**	**Median**	**Range**
Pelvic Angle *(degrees)*	10.1 (4.8)	10.0	0.0-19.0	12.2 (4.3)	12.3	4.0-21.0
Hip Anteversion *(degrees)*	8.3 (3.9)	9.2	1.0-16.7	13.0 (5.3)	10.7	4.3-27.7
Quadriceps Angle *(degrees)*	10.9 (4.3)	10.5	1.0-21.0	14.8 (6.2)	13.7	5.3-29.0
Tibiofemoral Angle *(degrees)*	10.2 (2.0)	10.3	5.0-13.0	11.2 (2.0)	10.7	7.7-15.3
Genu Recurvatum *(degrees)*	2.9 (2.8)	2.0	-1.0-10.0	4.7 (4.4)	3.2	-1.3-14.3
Navicular Drop *(millimeters)*	5.9 (3.8)	6.3	-3.3-14.3	7.4 (6.0)	6.5	-4.0-25.7
Hip ABD Torque *(Nm/kg)*	0.73 (0.22)	0.70	0.37-1.33	0.65 (0.15)	0.64	0.42-1.04
Hip EXT Torque *(Nm/kg)*	4.05 (0.92)	4.14	2.25-5.80	2.86 (0.81)	2.66	1.87-4.65
Hip Adduction *(degrees)*	10.2 (9.9)	11.7	-14.0-25.4	12.5 (11.0)	12.4	-15.3-35.5
Hip Internal Rotation *(degrees)*	-2.2 (6.4)	-1.3	-16.4-9.2	-2.4 (5.4)	-2.3	-9.6-12.8
Knee Valgus *(degrees)*	-2.7 (7.6)	-4.0	-14.1-17.0	2.5 (7.7)	3.7	-23.5-16.0
Knee External Rotation *(degrees)*	1.1 (4.5)	1.4	-8.6-12.0	4.2 (7.1)	4.1	-9.8-20.2

Table 13. Results of MANOVA Multivariate Tests

Dependent Variable	Mean Square	F	Sig.	Effect Size
Pelvic Angle	68.98	3.35	.073	0.47
Hip Anteversion	322.79	14.89	.000*	1.01
Quadriceps Angle	226.85	7.98	.006*	0.74
Tibiofemoral Angle	16.71	4.33	.042*	0.54
Genu Recurvatum	45.65	3.35	.072	0.48
Navicular Drop	32.27	1.29	.261	0.30
Hip Abduction Torque	0.09	2.48	.121	0.41
Hip Extension Torque	21.39	28.46	.000*	1.38
Hip Adduction Excursion	79.37	.73	.396	0.22
Hip Internal Rotation Excursion	0.93	.026	.871	0.04
Knee Valgus Excursion	396.39	6.77	.012*	0.67
Knee External Rotation Excursion	137.38	3.92	.052	0.52

* significant difference between sex

CHAPTER V

DISCUSSION

This study examined the relationships between static lower extremity posture, hip strength and dynamic alignment of the lower extremity during a single leg squat in males and females. The primary findings were that lower extremity postures associated with greater hip anteversion and tibiofemoral angles predicted decreased hip abduction and extension strength respectively, which in turn predicted decreased hip internal rotation and greater knee valgus and external rotation during the single leg squat. Additionally, directs relationships were noted between lower extremity posture and dynamic knee alignment, with greater hip anteversion and genu recurvatum predicting greater knee external rotation, and greater hip anteversion and navicular drop predicting greater hip internal rotation. These findings indicate that static lower extremity postures characterized by greater valgus alignment offers one explanation for decreased hip muscle strength. Further, these findings provide empirical data to support previous theories that greater static LEP and decreased hip muscle function, independently and in combination, contribute to greater hip and knee joint angles during functional activities.

Other findings of this study were that females had greater LEP characteristics of hip anteversion, quadriceps angle and tibiofemoral angle, decreased hip extension strength, and greater dynamic knee valgus compared to males. These sex differences, along with the identified relationships among them, provide a reasonable explanation for why females have been found to land with greater functional valgus collapse (Ford et al., 2003; Hewett et al., 2004; Lephart et al., 2002; Zeller et al., 2003), a position found to be predictive of ACL injury (Hewett et al., 2005). The following discussion will focus on the specific relationships and sex differences observed among these variables, followed by a discussion of clinical implications of the findings and directions for future research.

Effects of Static Lower Extremity Posture on Hip Strength

Proper functioning of the posterior-lateral hip musculature is essential to providing proximal stability for lower extremity motion during functional activities. Previous studies have observed a relationship between decreased neuromuscular function of the hip muscles in those with low back pain and lower extremity injuries (Beckman & Buchanan, 1995; Brindle et al., 2003; Bullock-Saxton, 1994; Friel et al., 2006; Ireland et al., 2003; Jaramillo et al., 1994; Nadler et al., 2000). While it remains unknown if decreased hip function actually contributed to these injuries, the reasons to explain the decreased hip function have also not been clearly identified.

Limited studies have suggested that differences in select LEP characteristics, in particular increased hip anteversion, may contribute to changes in the force and activation of the posterio-lateral hip musculature (Merchant, 1965; Nyland et al., 2004). Collectively these findings suggest that individuals who have increased hip anteversion will require increased force production to control the hip and pelvis. The limitation with these studies is that only one or select lower extremity posture characteristics were examined. Therefore, it was hypothesized that static postures characterized by greater pelvic angle, hip anteversion, quadriceps angle, tibiofemoral angle and genu recurvatum would predict decreased hip strength. The hypothesis was partially supported as greater hip anteversion predicted decreased hip abduction torque, and greater tibiofemoral angle predicted decreased hip extension torques.

Hip Anteversion and Hip Abduction Torque

The potential for increased hip anteversion to lead to decreased hip abduction strength can be attributed to changes in the length-tension and orientation of the G_{med} muscle. The gluteus medius is the primary abductor of the hip (Inman, 1947; Kumagai et al., 1997; Moore, 1992) and attaches proximally along the outer edge of the iliac crest and is fan-shaped spanning the iliac crest from the anterior superior iliac spine to the posterior superior iliac spine (Gottschalk et al., 1989). The muscle tapers into a strong tendon and attaches distally on the anterior-superior portion of the greater trochanter (Gottschalk et al., 1989). Greater hip anteversion would displace the distal attachment of the G_{med} (greater trochanter) more anteriorly, which has been shown to result in the muscle functioning more as an internal rotator of the hip (Delp et al., 1999). This occurs as the internal rotation moment arm of the anterior portion of the G_{med} increases while the abduction and external rotation moment arms of the middle and posterior portions decrease and switch towards internal rotation (Delp et al., 1999; Dostal & Andrews, 1981; Dostal et al., 1986). In this position, the pure hip abduction capabilities of the muscle are likely diminished. This is further supported by research that has demonstrated decreased sEMG activation amplitude of the G_{med} in those with increased relative hip anteversion (Nyland et al., 2004).

Tibiofemoral Angle and Hip Extension Torque

The primary extensor of the hip is the G_{max} which attaches proximally along the posterior gluteal line of the ilium, dorsal surface of the sacrum and coccyx, and the sacrotuberous ligament. It slopes inferior-laterally at a 45° angle across the ischial tuberosity and attaches distally into the superficial fibers of the iliotibial tract and the gluteal tuberosity of the femur (Kendall et al., 1993; Moore, 1992). No previous studies were identified that directly examined the relationship between greater tibiofemoral angle and decreased hip extension strength. Theoretically, increased hip adduction (associated with increased tibiofemoral angle) would displace the distal attachment of the G_{max} (greater trochanter) more inferiorly, which has the potential to decrease the force producing capabilities of the G_{max}. This is supported by a previous study where the moment arms of the posterior portions of the G_{max} were reduced during sagittal and rotational motions at the hip that essentially displaced the greater trochanter more anterior and inferior (Delp et al., 1999). Considering these changes in G_{max} function were more specific to sagittal and rotational changes in hip position, this would also support the near significant relationship ($t = 1.95$) noted between greater hip anteversion and decreased hip extension strength, as anterior displacement of the greater trochanter with greater hip anteversion would also decrease the force producing capabilities of the G_{max} muscle.

Summary

Static postures of the hip and knee angles were found to be predictive of posterio-lateral hip strength where greater hip anteversion and greater tibiofemoral angle predicted decreased hip abduction and hip extension torques, respectively. These relationships can be attributed to changes in the length-tension and orientation of the G_{med} and G_{max} muscles. Collectively, the G_{med} and G_{max} function to provide proximal stability for lower extremity motion during functional activities. Increased static LEP characteristics would reduce the muscular efficiency and force producing capabilities of these muscles and make stabilization of the hip and pelvis more difficult during functional activities. These findings support the possibility that greater LEP characteristics may be an underlying mechanism for reduced hip strength, potentially leading to greater dynamic motion during functional activities (Ford et al., 2003; Griffin et al., 2000; Hewett et al., 2004; McClay Davis & Ireland, 2003; Zeller et al., 2003).

These collective findings provide clinicians with a simple and cost effective tool to potentially identify those with decrease hip strength. Measurement of LEP characteristics can be clinically performed efficiently and with an acceptable level of reliability (Shultz, Nguyen et al., 2006). This provides clinicians with an avenue for screening for those who might have faulty postures leading to hip strength deficits, thus allowing them to better focus their detailed strength evaluations and intervention strategies on those individuals who may be at greatest risk.

Effects of LEP and Hip Strength on Functional Valgus Collapse

Functional valgus collapse of the knee, characterized by adduction and internal rotation of the hip and knee valgus during dynamic activities has been observed as a common mechanism of ACL injury (Ireland, 1999; Olsen et al., 2004) and found to be predictive of ACL injury risk (Hewett et al., 2005). Decreased function of the posterio-lateral hip musculature that are responsible for stabilizing the pelvis and maintaining proper hip and knee alignment, has been postulated as a potential contributor to this functional valgus collapse (Ford et al., 2003; Griffin et al., 2000; McClay Davis & Ireland, 2003; Zeller et al., 2003).

In addition to decreased function of the hip musculature, LEP has also been proposed as an intrinsic risk factor for ACL injury (Griffin et al., 2000; Hutchinson & Ireland, 1995; Ireland, 1999; Ireland et al., 1997; McClay Davis & Ireland, 2003). Retrospective studies have reported greater pronation, pelvic angle and genu recurvatum in those individuals that have suffered ACL injury (Beckett et al., 1992; Hertel et al., 2004; Loudon et al., 1996; Woodford-Rogers et al., 1994) These and other postural characteristics that increase static hip and knee angles may predispose individuals to increased inward collapse of the knee during functional activities.

This study was designed to test both of these theories by examining the relationships between lower extremity posture and neuromuscular function of the hip with functional valgus collapse. Based on prevailing theories, it was hypothesized that greater static hip and knee valgus (greater pelvic angle, hip anteversion, quadriceps angle, tibiofemoral angle, genu recurvatum) and decreased hip strength (abduction and extension) would predict greater joint motion during the single leg squat.

Direct Effects of LEP on Functional Valgus Collapse

Direct relationships were noted between greater hip anteversion and genu recurvatum with greater knee external rotation, and between greater navicular drop and hip anteversion with greater hip internal rotation. Greater hip anteversion was common to both of these relationships, predicting greater knee external rotation and knee hip internal rotation, both of which are considered components of functional valgus collapse and thought to contribute to the position of no return (Ireland, 1999). This direct relationship seems logical as greater hip anteversion essentially results in femoral internal rotation and contributes to a compensatory increase in knee external rotation (Hvid & Andersen, 1982).

These observed relationships suggest that postural characteristics may directly influence dynamic hip and knee angles during functional activities and may offer a potential mechanism by which previous studies have found greater navicular drop and genu recuvatum to be associated with ACL injury (Beckett et al., 1992; Hertel et al., 2004; Loudon et al., 1996; Woodford-Rogers et al., 1994). While sex differences have been rarely observed in navicular drop and genu recurvatum, their relationship with hip anteversion in predicting these motions emphasizes the need to consider the alignment of the entire lower extremity when examining their relationship with injury risk.

Direct Effect of Hip Strength on Functional Valgus Collapse

The hypothesized relationship between hip strength and functional valgus collapse was partially supported as decreased hip abduction and extension strength were found to be significant predictors of several characteristics of functional valgus collapse during a single leg squat. Specifically, decreased hip abduction torque predicted greater knee external rotation while decreased hip extension torque predicted greater knee valgus during a single leg squat. These relationships support current theories that decreased hip strength and control may decrease proximal stability for lower extremity motion, resulting in an inability to maintain a neutral alignment of the hip and knee during single limb weight bearing activities. (Ferber et al., 2003; Lephart et al., 2002; Malinzak et al., 2001; Zeller et al., 2003) These findings are in agreement with the one other study that examined the relationship between hip strength and lower extremity motion during a functional task. Padua et al (Padua et al., 2005) examined the relationship between hip strength (measured by a hand-held dynamometer) and joint kinematics during a drop-jump task in 63 males and 54 females and reported decreased G_{med} and G_{max} strength was related to greater knee valgus at initial contact and greater peak knee valgus. These collective findings demonstrate the importance of hip strength in controlling dynamic motion of the knee and their role in preventing dynamic malalignments that are predictive of knee injury.

Indirect Effects of LEP on Functional Valgus Collapse

While direct relationships were observed between LEP and functional valgus collapse, it is unclear from these data alone if static LEP directly predisposes individuals to functional valgus collapse or whether these postural effects act through resulting biomechanical changes (i.e. decreased hip strength) to increase dynamic hip and knee malalignments. The path analysis model used in this study also examined the indirect relationships between LEP and functional valgus collapse, by way of their effects on posterio-lateral hip strength. Interpretation of these indirect relationships indicate that greater hip anteversion led to decreased hip abduction torque which in turn led to greater dynamic knee external rotation. Similarly, greater tibiofemoral angle led to decreased hip extension torque, leading to greater dynamic knee valgus. These findings suggest a “sequential” relationship that has not been previous addressed, and provide empirical evidence to support proposed theories that LEP may alter hip muscle function, leading to reduced hip control and increased dynamic lower extremity malalignments during functional activities (Ford et al., 2003; Griffin et al., 2000; Hewett et al., 2004; McClay Davis & Ireland, 2003; Nyland et al., 2004; Zeller et al., 2003).

Direct and Indirect Effects of Hip Anteversion on Functional Valgus Collapse

A valgus posture which represents the interrelationships among LEP characteristics of pelvic angle, hip anteversion, quadriceps angle, tibiofemoral angle and genu recurvatum was originally proposed as a predictor variable. This was based on the assumption that these clinical measures of static hip and knee alignment would be related in the same manner as previously identified in a cohort of 100 physically active males and females (Nguyen & Shultz, In Review). However, preliminary analyses determined that relationships among the variables in the current subset of subjects were insufficient to collapse the variables into a single valgus postural factor for this study. Statistically, the lack of co-variation among the variables in the current set of subjects may be insufficient to identify these postural relationships. This may in part be due to a smaller sample size compared to the previous study (60 vs. 100), as this may have decreased the power when using a factor analysis model. Further, when the factor loadings of these LEP characteristics in the previous study were examined, the strength of the relationship between each LEP characteristic and the valgus posture factor were not equal across all variables. Hip anteversion had the highest loading ($\alpha = .810$) followed by quadriceps angle ($\alpha = .716$), tibiofemoral angle ($\alpha = .520$) and genu recurvatum ($\alpha = .520$), with pelvic angle having the least amount of loading ($\alpha = .457$) on the valgus posture. This suggests that certain LEP characteristics may contribute more to a valgus posture than others. Therefore, the effects of LEP characteristics of pelvic angle, hip anteversion, quadriceps angle, tibiofemoral angle and genu recurvatum were examined independently rather than as factor loadings on the latent variable of valgus posture.

Interestingly, hip anteversion, which was observed to have the highest factor loading in the previous study that examined interrelationships among LEP characteristics (Nguyen & Shultz, In Review), was also found to be the most consistent predictor of hip strength and functional valgus collapse during a single leg squat. As previously stated, greater hip anteversion results in increased femoral internal rotation, which anatomically places the knee in a more “at risk” position, as it was found to directly predict greater dynamic hip internal rotation and knee external rotation during a single leg squat. In addition, the potential for increased hip anteversion to change the distal attachment of the G_{med} and G_{max} muscle leading to decreased hip abduction and extension strength may also lead to these undesirable joint angles during dynamic motion and ACL injury. Hence, greater hip anteversion may be an important risk factor for ACL injury, given both its direct and indirect effects on functional valgus collapse. However, prospective study designs are needed to determine the true predictive ability of hip anteversion in identifying those at increased risk for ACL injury.

Relationship between Hip Strength and Activation

The purpose of accounting for G_{med} and G_{max} activation in the path analysis was to better clarify the relationship between hip strength and functional valgus collapse by accounting for variations in the level of muscle activation among subjects that may in itself explain differences in functional valgus collapse. While studies have examined activation of the hip musculature during functional activities such as single leg landings and single leg squatting, kinematic data were not collected (Zazulak et al., 2005) and/or hip strength was not reported (Zazulak et al., 2005; Zeller et al., 2003). From these studies, the relationship among posterio-lateral hip muscle function and dynamic joint motion remains unclear. In theory, greater hip muscle activation would be necessary to successfully perform a desired motion in the presence of reduced hip muscular strength. The significant negative relationships observed between hip abduction torque and G_{med}

activation (r = -.275), and between hip extension torque and G_{max} activation (r = -.612) in the current study confirmed that greater posterio-lateral hip muscle activation was required in those subjects with decreased hip strength in order to successfully perform the single leg squat. This may explain the increased G_{max} activation reported by Zeller et al. (2003) in females who still demonstrated greater hip adduction and knee valgus than males. However, the negative relationship between hip abduction torque and G_{med} activation observed in the current study does not explain the decreased G_{med} activation in females reported by Zeller et al. (Zeller et al., 2003) A plausible explanation for these contrasting findings may be differences in the methods of performing the single leg squat. In the current study, subjects were instructed to maintain an upright posture in an effort to control flexion at the trunk in the current study. Conversely, it does not appear that trunk motion was controlled in the previous study as subjects were instructed to perform a single leg squat down as far as possible without losing their balance. Because trunk flexion was allowed to occur, the decreased G_{med} activation they reported may have been observed even if decreased hip abduction strength was present. This is based on previous findings where flexion of the trunk was found to decrease G_{med} activation, and may require less force to maintain a single leg stance position (Schmitz et al., 2002). Therefore, while decreased G_{med} activation may not be reflective of hip abduction strength as the trunk flexes, decreased hip abduction strength may require more G_{med} activation with the trunk more upright, a position observed during ACL injury (Ireland, 1999).

While assessment of hip muscle activation measured by sEMG may not be readily available to clinicians and may be interpreted to have little clinical meaning, the results from the current study provides a potential link between research and clinical interpretation as future studies will continue to use sEMG as a measure of hip neuromuscular control. More specifically, the inverse relationship observed between hip muscle strength and activation suggests that an increase in gluteal muscle activation may or may not be in itself indicative of better hip control, depending on the actual torque producing capabilities of the muscles.

Sex Differences in Potential Risk Factors for ACL Injury

Decreased function of the posterio-lateral hip musculature has been postulated as a potential reason for why females have a greater tendency towards functional valgus collapse (Ford et al., 2003; Griffin et al., 2000; McClay Davis & Ireland, 2003; Zeller et al., 2003). However, the underlying causes for this dysfunction and greater prevalence in females have received little attention**.** Based on this unknown, sex differences in lower extremity posture, which have also been proposed as an ACL injury risk factor, were examined for their effects on posterio-lateral hip muscle function and their collective contribution to dynamic limb alignment between males and females. Given the greater prevalence of functional valgus collapse in females, and the theoretical relationships between LEP, posterio-lateral hip strength and dynamic knee valgus, it was hypothesized that females would have 1) a static posture characterized by greater hip and knee valgus,

2) decreased normalized hip strength, and therefore 3) demonstrate greater functional valgus collapse during the single leg squat compared to males.

This hypothesis was in large part supported as females compared to males had greater hip anteversion, quadriceps angle and tibiofemoral angle, decreased hip extension torque, and greater dynamic knee valgus and external rotation during the single leg squat. Trends were also noted for females having greater pelvic angle and genu recurvatum, and decreased hip abduction torque as these variables also approached significance. The moderate effect sizes associated with these variables (.41-.52) suggest that the sample size was too small to yield sufficient statistical power to reach significance. Box plots for all variables by sex are presented in Appendices D_1-D_3. The following will discuss the sex differences observed and their contributions to the current body of literature.

Sex Differences in Lower Extremity Posture

The mean values for LEP characteristics (Table 14) are within the range of normal values reported previously for healthy adults using similar clinical measurement methods (Aglietti et al., 1983; Alviso et al., 1988; Beckett et al., 1992; Evans et al., 2003; Gajdosik et al., 1985; Gilliam et al., 1994; Guerra et al., 1994; Horton & Hall, 1989; Hsu et al., 1990; Livingston & Mandigo, 1997; Nguyen & Shultz, In Press; Picciano et al., 1993; Scerpella et al., 2005; Trimble et al., 2002; Woodland & Francis, 1992). The current findings of greater LEP characteristics in females compared to males supports study hypothesis 3, and the magnitude of observed sex differences for hip anteversion (~5º), Q-angle (~4º), tibiofemoral angle (~1º), pelvic angle (~2º) and genu recurvatum

(~2°) are consistent with other sufficiently powered studies that examined sex differences using similar measurement methods (Aglietti et al., 1983; Braten et al., 1992; Guerra et al., 1994; Hertel et al., 2004; Horton & Hall, 1989; Hsu et al., 1990; Livingston & Mandigo, 1997; Nguyen & Shultz, In Press; Prasad et al., 1996; Trimble et al., 2002; Woodland & Francis, 1992). While sex differences in measures of pelvic angle and genu recurvatum were not statistically significant, there was a trend toward greater pelvic angle (p=.073) and genu recurvatum (p=.073) in females compared to males, which were associated with medium effect sizes (.47 and .48, respectively). This suggests that meaningful sex differences would have been observed with a larger sample size.

These sex differences in LEP characteristics in the hip and knee may in part explain why females have been consistently found to land and cut with greater valgus angles and moments compared to males (Ford et al., 2003; Hewett et al., 2004; Hewett et al., 2005; Lephart et al., 2002; Zeller et al., 2003). This is based on the current findings where greater angles in several of these LEP were found to be predictive of greater hip internal rotation and greater knee external rotation. Therefore, greater static malalignments of the hips and knees in females, and their relationship to components of functional valgus collapse, offer a plausible explanation for the greater risk of ACL injury in females.

While the absence of a sex difference in mean navicular drop values was expected and is consistent with previous studies in the adult population (Beckett et al., 1992; Hertel et al., 2004; Moul, 1998; Trimble et al., 2002), there is evidence that this variable may interact with other LEP characteristics that do differ between sex, to contribute to functional valgus collapse. This is supported in the current study as greater navicular drop and hip anteversion predicted greater hip internal rotation during the single leg squat. Previous studies also support this potential interactive effect as the combinations of navicular drop with pelvic angle (Hertel et al., 2004) and navicular drop with genu recurvatum (Loudon et al., 1996) were reported to explain the greater risk of knee injuries in females. However, while the findings of the current study identified a moderate positive correlation between greater navicular and genu recurvatum ($r = .406$), these variables in combination did not predict reduced hip strength or increased dynamic malalignment during the single leg squat. More work is needed to clarify the potential interactive effects of navicular drop with other alignment variables, and their potential to influence dynamic knee alignment and ACL injury risk in females.

Sex Differences in Hip Strength

Consistent with greater hip anteversion and knee valgus angles in females, and the relationships previously noted between these static postural variables and posterio-lateral hip function (i.e. Hypotheses 1 and 2), females produced less hip extension torque. These findings are consistent with Cahalan et al. (1989) who assessed isometric hip extension strength in a standing position while supported in a body stabilization frame with the hip in 90° of flexion. While direct comparison of mean values between studies is difficult due to methodological differences, the relative magnitudes of the differences are consistent. Specifically, Cahalan et al. (1989) reported that females on average produced approximately 1.6 times less average extension peak torque than males (126 Nm vs. 204

Nm) where females in the current study were observed to produce approximately 1.4 times less normalized peak extension torque in a supine position.

While not statistically significant, there was also a trend toward decreased hip abduction torque in females compared to males. Males demonstrated only a modest increase in strength above females (1.12), yet previous studies have found that males were able to produce over 1.2-1.5 times more hip abduction force/torque compared to females (Bohannon, 1997; Cahalan et al., 1989; Leetun et al., 2004; Murray & Sepic, 1968). One reason for this discrepancy may be in the different methods used for measurement. In the current study, isometric hip abduction torque was measured in a functional position while standing on the dominant stance limb. In previous studies, subjects were tested in non-weight bearing (Leetun et al., 2004; Murray & Sepic, 1968) or partial weight bearing (Cahalan et al., 1989) positions. These positions may not adequately represent the force producing capabilities of the hip abductors compared to a more functional position where the muscles have additional requirements to stabilize the pelvis in order to produce a force. This latter test position was chosen as pilot testing indicated that G_{med} activation of the stance leg was greater than the non-stance leg during the MVIC in the functional. Based on these methodological differences, more work is needed to determine the most effective test positions for determining the most reliable and valid measurement of abduction torque for comparison between and within subjects.

Based on the findings of reduced hip strength in females, and the relationships observed in the current study between decreased hip strength and greater knee external rotation and valgus during a single leg squat, reduced hip strength provides another plausible explanation for why females demonstrate greater dynamic knee angles and an increased risk of ACL injury.

Sex Differences in Functional Valgus Collapse

It was hypothesized that females compared to males would have greater functional valgus collapse characterized by greater hip adduction, hip internal rotation, knee valgus and knee external rotation during a single leg squat. This hypothesis was partially supported as females were observed to have greater knee valgus excursion and a trend toward greater knee external rotation ($P = .052$) during the single leg squat, however no sex differences were present in hip adduction and hip internal rotation.

During the single leg squat, females tended to move towards an average of 2.5º knee valgus while males tended to move towards 2.7º of knee varus. This is in contrast to one other study that examined joint motion during a single leg squat and observed significantly greater hip adduction in females, but not in knee valgus excursion (Zeller et al., 2003). In fact, the previous study reported that both males and females moved towards a varus knee alignment but noted that females initially started in a more valgus alignment while males initially started in a varus alignment. Conversely, both males and females in the current study demonstrated initial angles that were close to neutral alignment (0.6º males, 0.9º females), with males moving into more varus as females moved into more valgus. While this study examined a sample of 60 subjects (30 males, 30 females), the previous study was based on a relatively small sample size of 18 subjects

(9 males, 9 females) which may not have been sufficiently representative of the population as a whole.

Sex differences in knee valgus excursion were also accompanied by approximately 3° greater knee external rotation excursion in females compared to males that was very close to significant ($p = .052$; effect size .52). While knee external rotation (tibial external rotation relative to the femur) has been identified as a characteristic of functional valgus collapse, studies examining sex differences in knee external rotation during functional activities are limited. One study that examined knee rotation during running observed that mean knee rotation values for both males and females moved from knee external rotation to internal rotation during the stance phase of running (Ferber et al., 2003). While the knee rotational patterns they observed during running were different than what was observed in the current study during the single leg squat, they were consistent in finding that females had approximately 2º greater peak knee external rotation angles compared to males. The authors suggested that increased hip internal rotation observed in female runners led to greater knee external rotation values. This compensatory increase in knee external rotation with hip internal rotation has been previously reported (Hvid & Andersen, 1982) and is consistent with the current findings of greater hip anteversion and decreased posterio-lateral hip strength in females, leading to greater femoral internal rotation and knee external rotation. Further, the combination of knee valgus and external rotation has been shown to impinge the ACL against the femoral intercondylar notch in computer simulated models (Fung & Zhang, 2003) and should be concerning components of functional valgus collapse as a mechanism for ACL injury.

In the current study, females were observed to have approximately 2º more average hip adduction but essentially no differences (0.2º) in hip internal rotation compared to males during the single leg squat. The absence of a sex difference in these joint excursions is in contrast to previous studies noting approximately 4º more hip adduction and hip internal rotation in females compared to males during more dynamic activities such as single leg landing (Lephart et al., 2002) and running (Ferber et al., 2003). While it is possible that the magnitude of sex differences in joint excursions may increase with the dynamics of the activity, further research is needed to examine the extent and consistency of sex differences in joint motions during a variety of functional activities.

Summary

In summary, and based on results supporting hypothesis 1 and 2, the collective sex differences observed in LEP, hip strength and dynamic knee alignment during the single leg squat indicate that females are more prone to postures that promote decreased posterio-lateral hip strength and increase knee valgus and external rotation. Although the amount of hip and knee rotation may be scalar to the dynamics of the task, more work is needed to determine if subjects would show consistent patterns of hip and knee valgus and rotation among various tasks, or whether these patterns are task specific. The rationale for examining joint motions during a single leg squat in the current study was that this task may in time allow clinicians to more accurately observe hip and knee joint excursion during a controlled, functional movement without the use of expensive motion analysis systems. Given the relationships noted between LEP, hip strength and these joint motion patterns, it would seem reasonable that these patterns might be consistent within an individual across tasks. Future work in this area is important, as it will help establish the validity of the single leg squat as a simple, cost effective screening tool that may be clinically useful to identify those who demonstrate at risk knee positions.

These findings add to the current body of literature by providing plausible explanations for why females have often been observed to have decreased hip strength compared to males (Bohannon, 1997; Cahalan et al., 1989; Leetun et al., 2004; Murray & Sepic, 1968), and support LEP as an underlying mechanism(s) for why females are more often observed to demonstrate a functional valgus collapse during functional tasks (Ford et al., 2003; Griffin et al., 2000; Hewett et al., 2004; McClay Davis & Ireland, 2003; Zeller et al., 2003). More work is needed to determine if the joint motion patterns during a single leg squat place the knee at greatest risk for injury, and whether females that demonstrate greater knee valgus during the single leg squat will also demonstrate greater knee valgus during other more dynamic activities is unknown. As previous studies have observed greater knee valgus in females during other more ballistic functional activities such as running, cutting and landing (Ferber et al., 2003; Ford et al., 2003; Ford et al., 2006; Hewett et al., 2004; Malinzak et al., 2001), this relationship is plausible, but has yet to be examined. Given the already established relationship between reduced hip strength (Leetun et al., 2004), functional valgus collapse (Hewett et al., 2005) and knee injury in females, these connections may in part explain the greater risk of ACL injury in females. However, this study did not specifically examine injury risk, and prospective studies are now needed to examine the extent to which LEP and hip strength in combination may predict actual injury risk.

Clinical Implications of Findings

While decreased hip strength was found to be a contributor to increased dynamic joint angles that are thought to be predictive of ACL injury, equipment and time demands make it difficult to readily assess hip strength. Therefore, it would seem clinically beneficial to be able to identify those at risk of injury in a large athletic population, using simple, clinical screening tools. Based on the findings of this study, clinical assessments of static LEP and dynamic knee and hip alignment during a controlled single leg squat may offer an efficient and clinically useful method for identifying those that demonstrate decreased hip strength.

The overall findings of this study revealed that greater LEP characteristics predicted decreased posterio-lateral hip strength. Further, greater LEP characteristics and decreased posterio-lateral hip strength, separately and in combination, predicted greater functional valgus collapse during the single leg squat. These collective findings provide clinicians with a simple and cost effective tool to identify those with decreased hip strength based on poor dynamic hip control. In addition, the findings that greater hip anteversion and tibiofemoral angle were predictive of decreased hip abduction and extension torque provides clinicians with possible contributors to the decreased strength, and another avenue for screening for those who might have hip strength deficits. While it may not be possible to correct the static postures leading to the strength deficits and dynamic malalignments, clinicians may be able to better focus their intervention strategies on individuals with these faulty postures. For example, if individuals who demonstrate an excessive valgus posture may also demonstrate greater functional knee valgus during a single leg squat, intervention strategies that focus on increasing strength and dynamic control of the posterior-lateral hip musculature may be warranted to counteract the effects of this faulty static posture.

The findings from this study are limited to the dominant stance limb of healthy, college age adults and should not be generalized to other populations. These findings are also limited to a controlled, functional single leg squat task performed in an upright position, and may not be representative of other more dynamic functional tasks such as landing, jumping and cutting. It is also important to emphasize that the values obtained for lower extremity posture characteristics using clinical measurement methods is representative of a single examiner with known measurement reliability.

Future directions for research should include continued studies to confirm whether the relationship between posture, hip strength and dynamic knee valgus are consistent across a variety of functional tasks, and prospectively examine whether these risk factors are truly predictive of ACL injury. In addition, continued examination of differences in lower extremity posture characteristics across older and younger individuals are needed to determine whether these postures change with maturity. This research will aid clinicians in determining the most appropriate time to initiate posterio-lateral hip strengthening programs towards reducing injury. The current study would also benefit from an understanding of the relationship between lower extremity posture and a more comprehensive assessment of other trunk and lower extremity muscles that control lower extremity motion. Finally, appropriate intervention strategies that target specific areas known to contribute to decreased dynamic control of the lower extremity are needed and evaluated for effectiveness. Continued work in these areas will help clinicians effectively identify those at greater risk for injury, and therefore help us develop intervention strategies to subsequently reduce the risk of ACL injury.

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APPENDIX A. Institutional Review Board Consent Form

UNIVERSITY OF NORTH CAROLINA AT GREENSBORO
CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM

Project Title: Relationships Between Lower Extremity Posture And Lower Extremity Kinematics With Posterior Lateral Hip Activation During A Single Leg Squat

Project Director: Anh-Dung Nguyen

Participant's Name: ______

PURPOSE OF THE RESEARCH:
The primary purpose of this study is to examine the relationship between knee joint motion, lower extremity posture, and hip muscle activation during a single leg squat.

DESCRIPTION AND EXPLANATION OF PROCEDURES:
In order to qualify for this investigation, you must have no previous history of surgery in the lower extremity, no history of lower extremity injury in the past 6 months, or any previous history that would affect the alignment or motion of your lower extremity joints (i.e. hip, knee or ankle). If you meet these criteria, you will be asked to attend one, 90-minute testing session. All testing will be performed in the Applied Neuromechanics Research Laboratory at UNCG. During the test session, you will be asked to complete a short physical activity questionnaire. We will record your height, weight, age, and we will measure the alignment of your hips, knees and legs. Your dominant leg will then be shaved and wiped with alcohol swabs in preparation for placement of two electrodes on the muscle on the side of your hip (gluteus medius) and the back of your hip (gluteus maximus). If you wish, a same-sex examiner will be available to apply the electrodes. The electrodes will be connected to wires that lead to a computer that measures muscle activity. With the electrodes attached, you will be asked to perform maximal contractions of the posterior and lateral hip muscles by turning your hip outward and by pushing your leg out to the side away from your body. These motions will be resisted by a dynamometer that will record the force of the contraction. Five trials will be recorded. You will then have 5 small position sensors (one on the mid back, low back, thigh, leg and foot) attached with double-sided tape and non-adhesive elastic tape. You will be asked to perform a one leg squat in a limited range of motion. Five trials will be recorded to measure the position of your hip and knee and the activation of your hip muscles during this squat task.

RISKS AND DISCOMFORTS:
There is a minimal chance you may suffer a muscle strain when you are being tested for muscle strength using a maximal contraction. To guard against this risk, you will

complete a 5 minute bike warm up and stretch prior to the strength test. If at any time you feel pain or discomfort, you should stop the contraction.

POTENTIAL BENEFITS:
There are no direct benefits to you for participating in this study. This study will provide data for future studies that may help us better understand the risk factors associated with injuries to the lower extremity.

CONFIDENTIALITY AND DATA MANAGEMENT:
The information that is obtained from this study will be handled confidentially. All consent forms will be maintained in a confidential file only accessible by the investigator. Your information will be assigned a code number. The list connecting your name to this number will be kept in a locked file. When the study is completed and the data have been analyzed, this list will be destroyed. Your name will not be used in any report. They will be kept in a file in a locked room for 3 years at which time they will be destroyed by shredding. All data will be stored on the principal investigators personal computer identified only by subject number. All data disks will be erased once all manuscripts of the data have been submitted and published for two years. A photocopy of this original consent form will be provided to you for your records.

CONSENT:
By signing this consent form, you agree that you understand the procedures and any risks and benefits involved in this research. You are free to refuse to participate or to withdraw your consent to participate in this research at any time without penalty or prejudice; your participation is entirely voluntary. Your privacy will be protected because you will not be identified by name as a participant in this project.

The University of North Carolina at Greensboro Institutional Review Board, which insures that research involving people follows federal regulations, has approved the research and this consent form. Questions regarding your rights as a participant in this project can be answered by calling Mr. Eric Allen at (336) 256-1482. Questions regarding the research itself will be answered by Anh-Dung Nguyen by calling 336-334-3039 or by Sandra Shultz by calling 336-334-3027. Any new information that develops during the project will be provided to you if the information might affect your willingness to continue participation in the project.

By signing this form, you are agreeing to participate in the project described to you by Anh-Dung Nguyen.

______________________________________ _______________

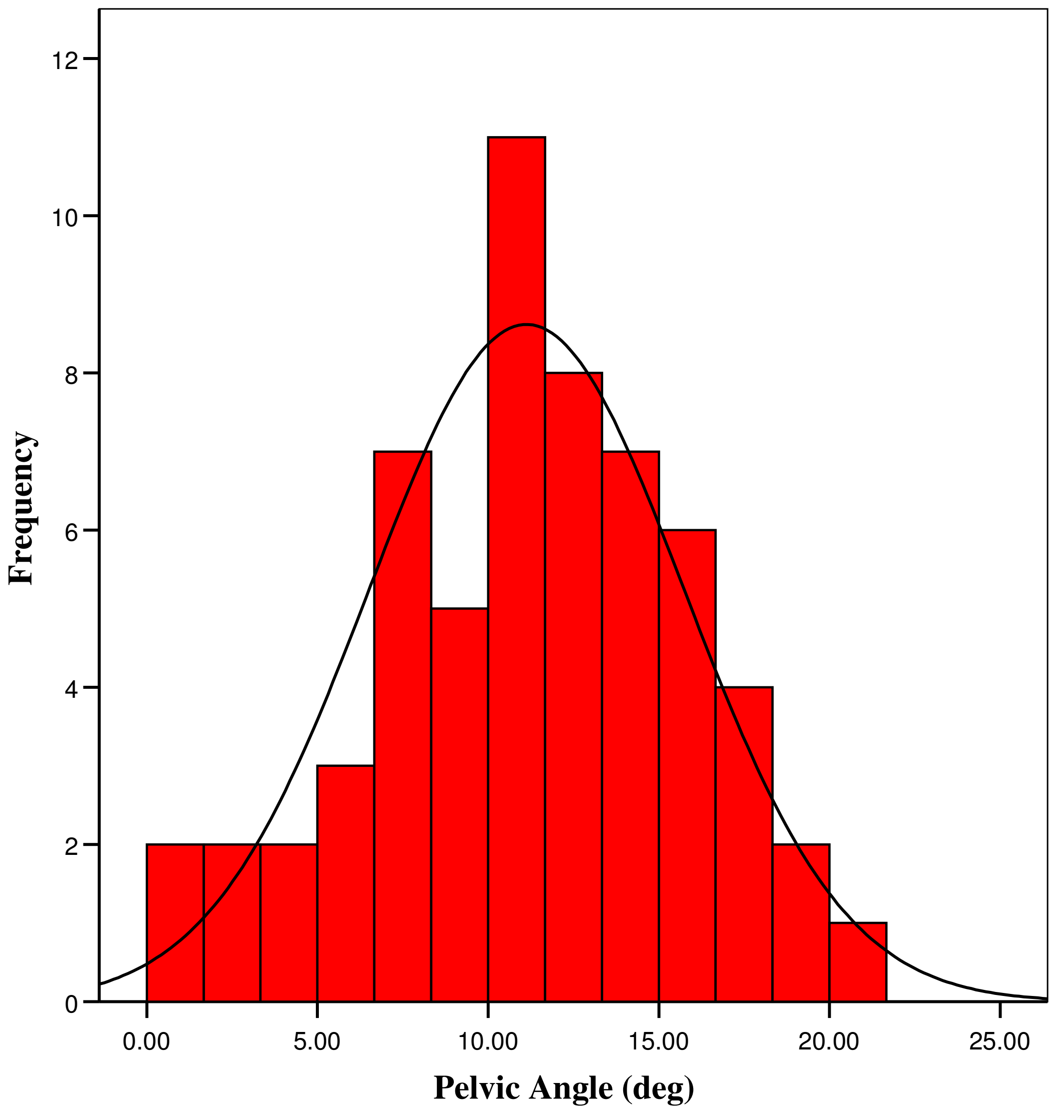
Participant's Signature* Date

APPENDIX B. Activity Rating Scale (Adapted from Marx et al, 2001)

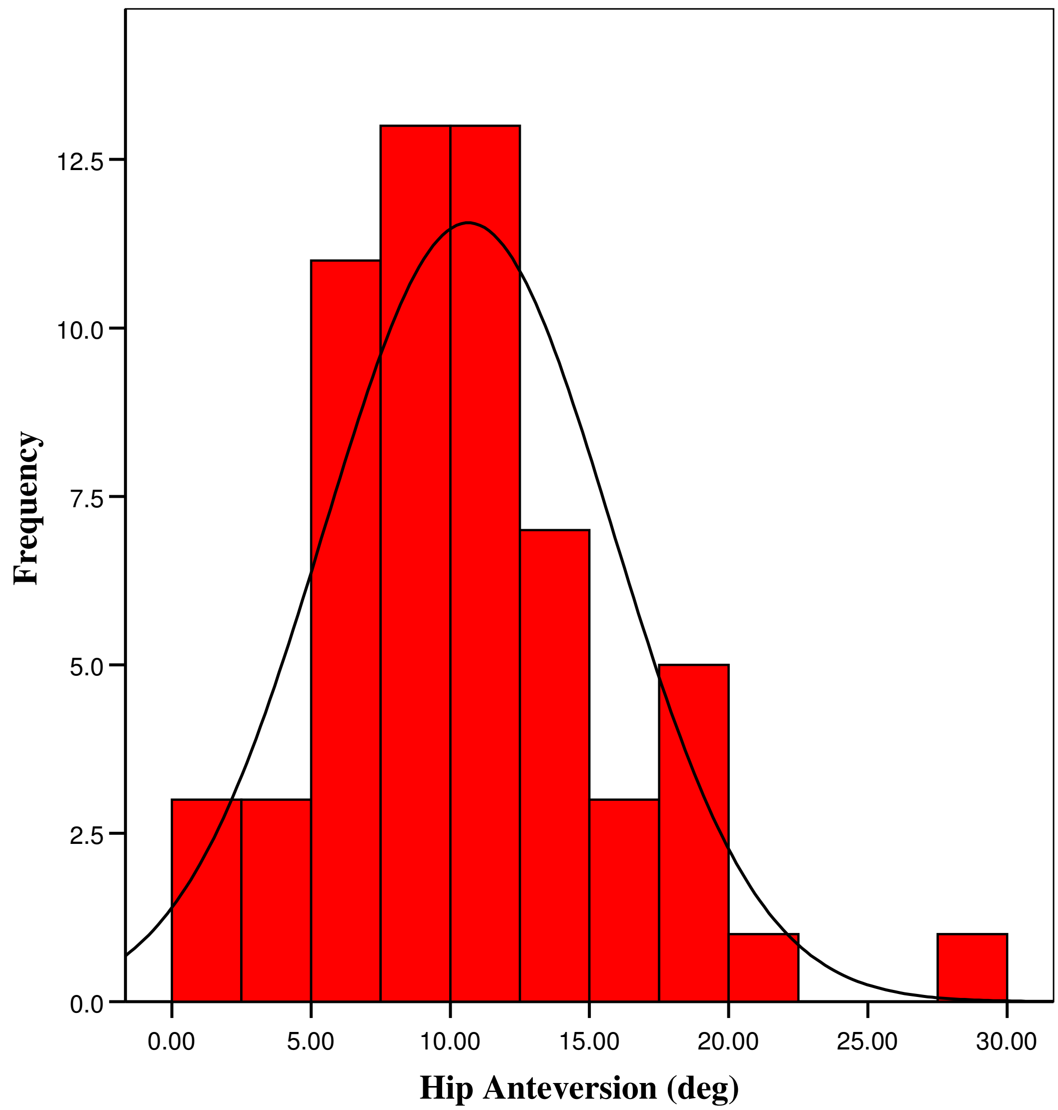
Please indicate how often you performed each activity in your healthiest and most active state, **in the past year.**

	Less than one time in a month	**One time in a month**	**One time in a week**	**2 or 3 times in a week**	**4 or more times in a week**
Running: running while playing a sport or jogging					
Cutting: changing directions while running					
Decelerating: coming to a quick stop while running					
Pivoting: turning your body with your foot planted while playing a sport; For example: skiing, skating, kicking, throwing, hitting a ball (golf, tennis, squash), etc.					

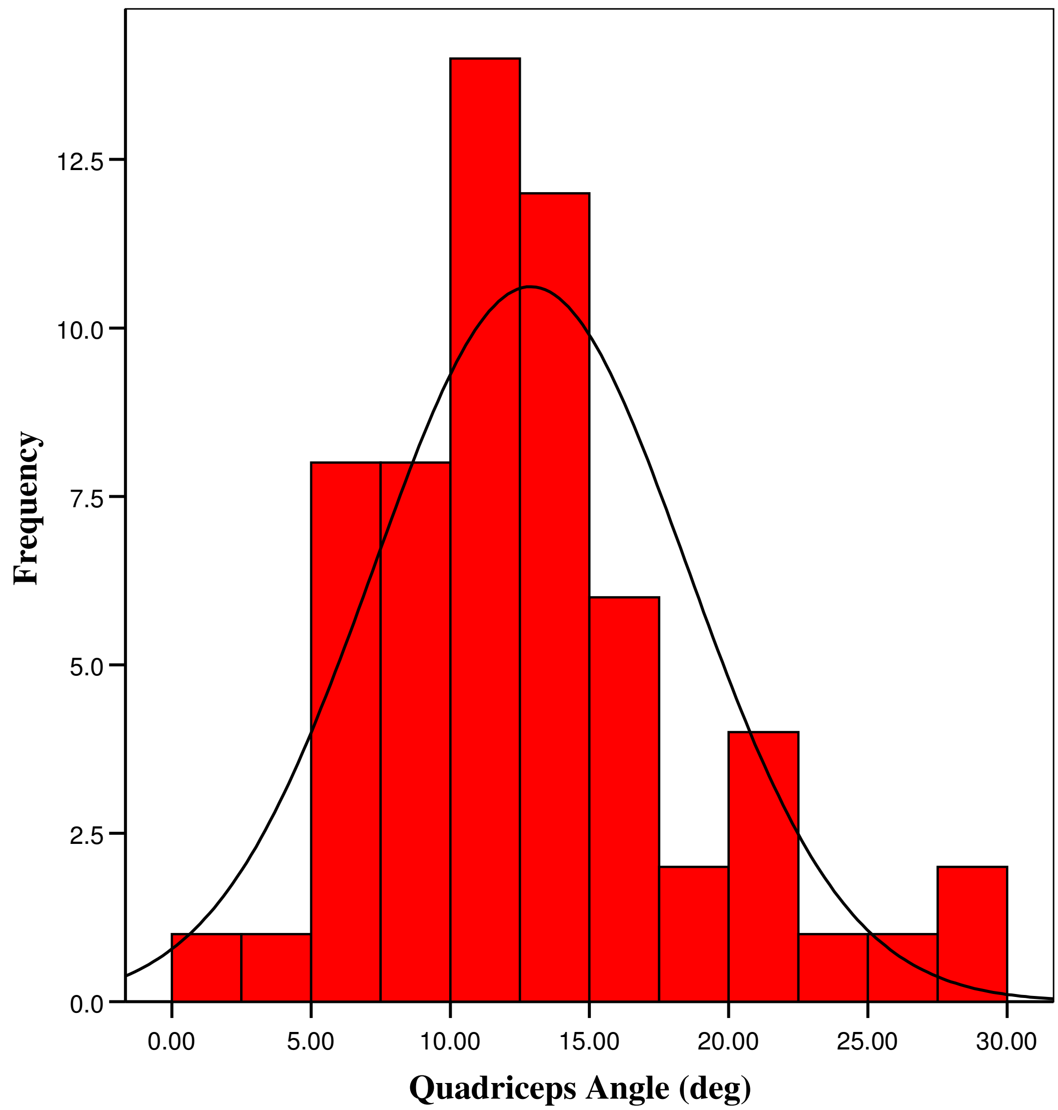
APPENDIX C_1. Pelvic Angle Histogram



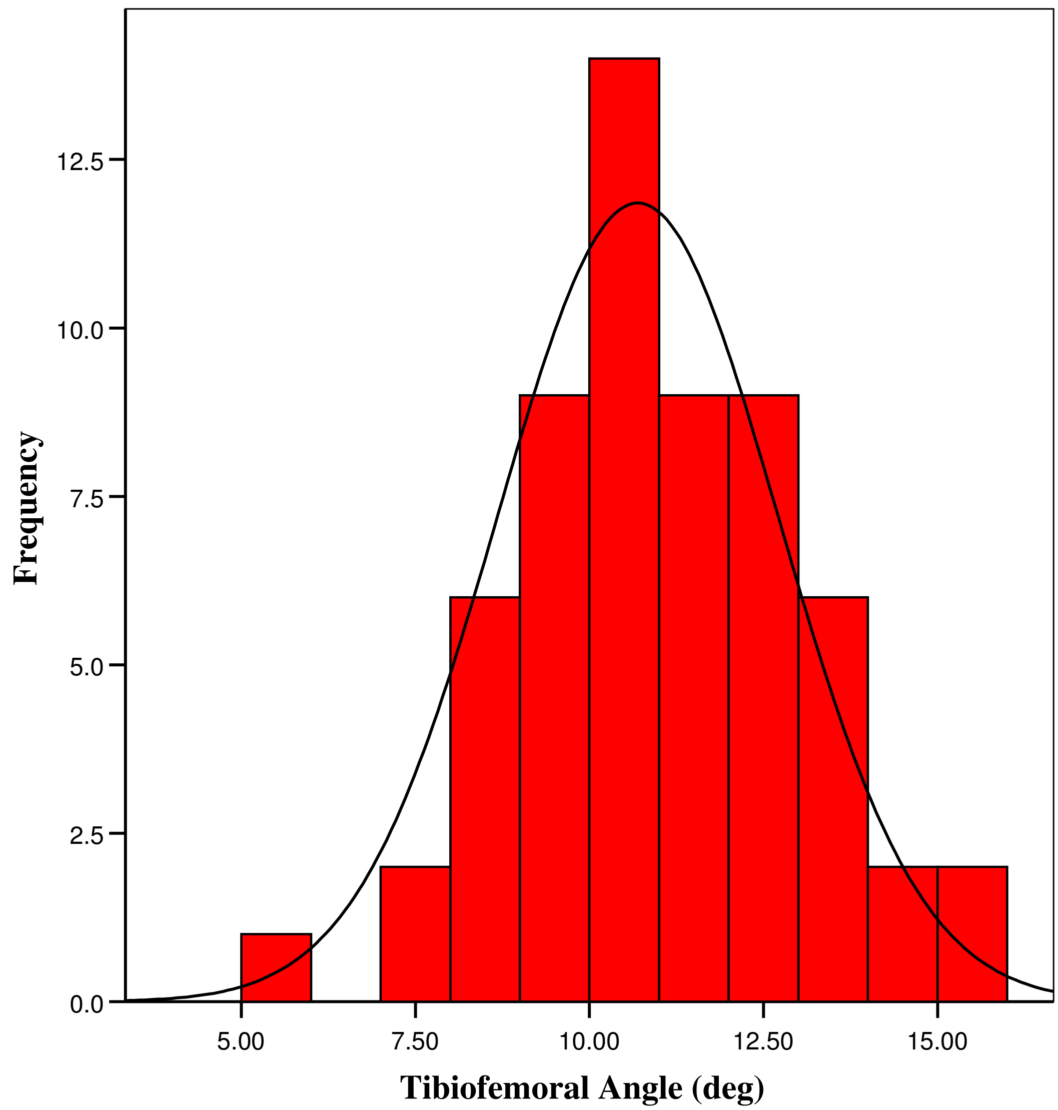
APPENDIX C_2. Hip Anteversion Histogram



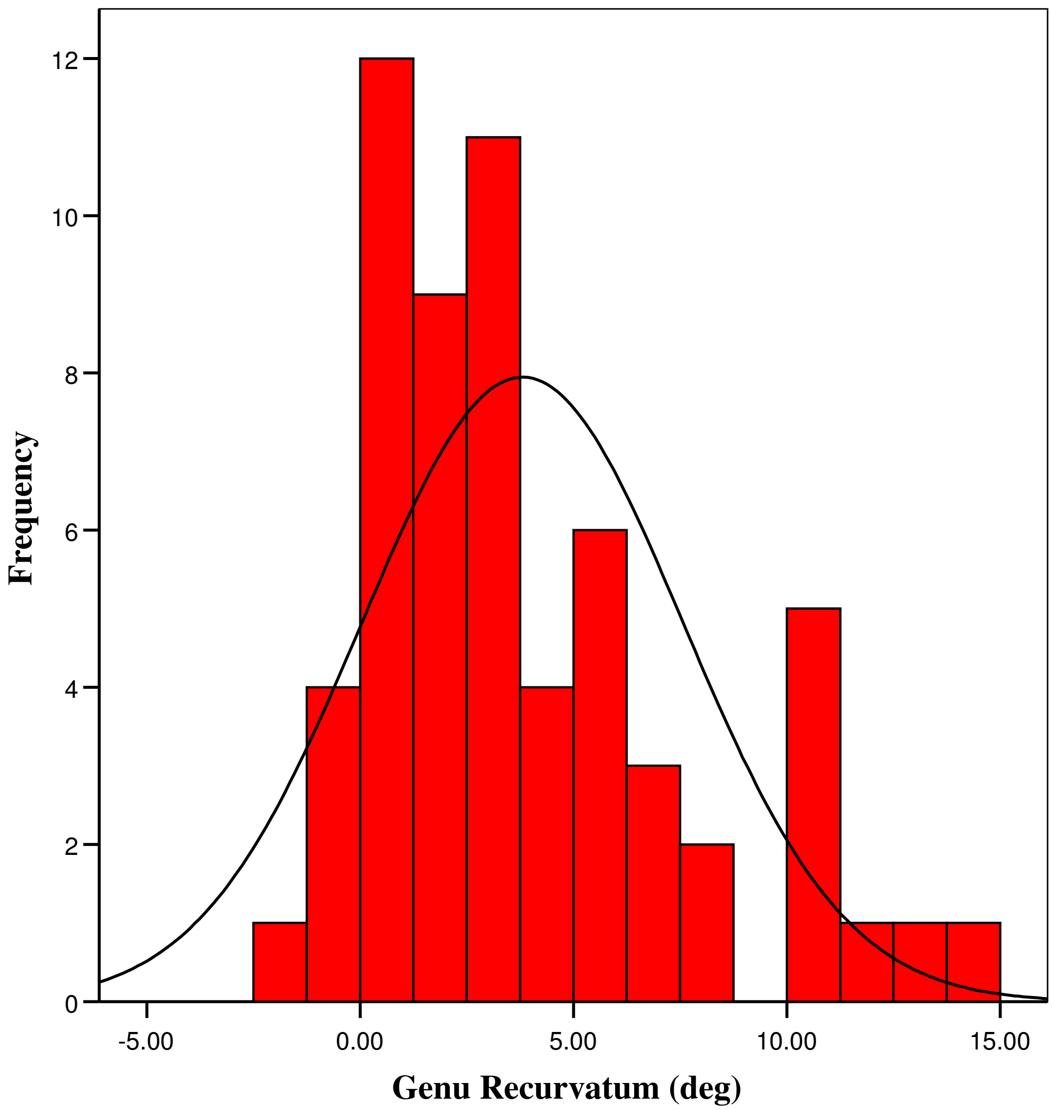
APPENDIX C_3. Quadriceps Angle Histogram



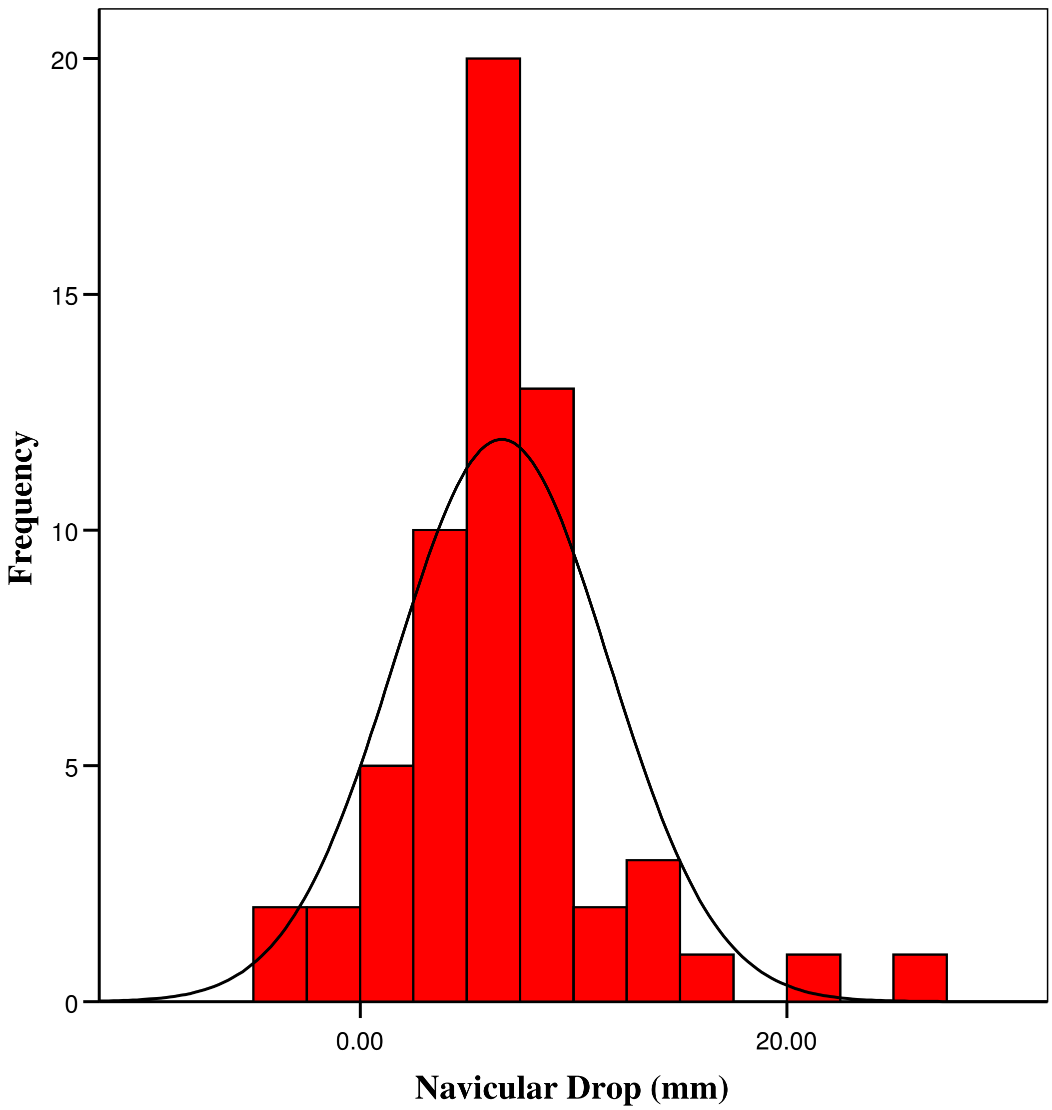
APPENDIX C_4. Tibiofemoral Angle Histogram



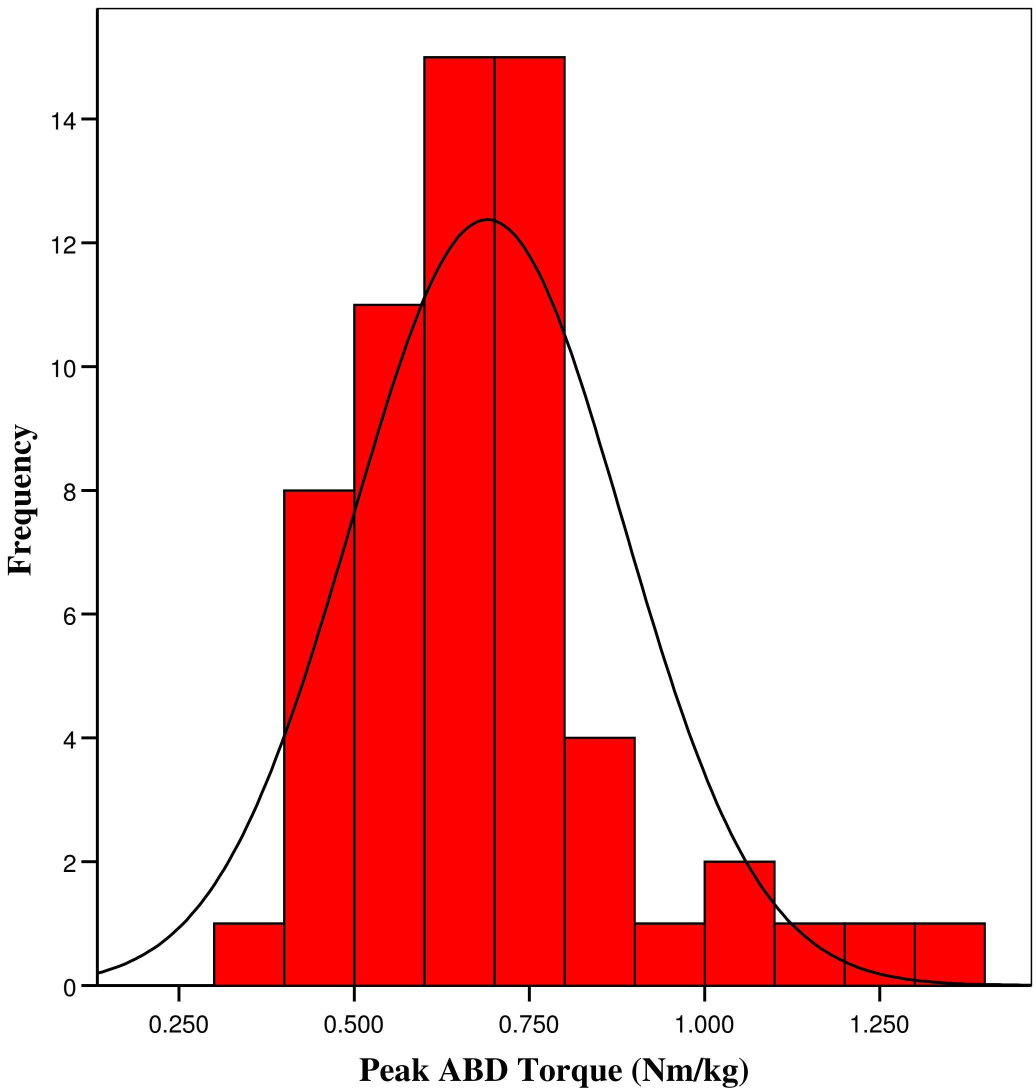
APPENDIX C_5. Genu Recurvatum Histogram



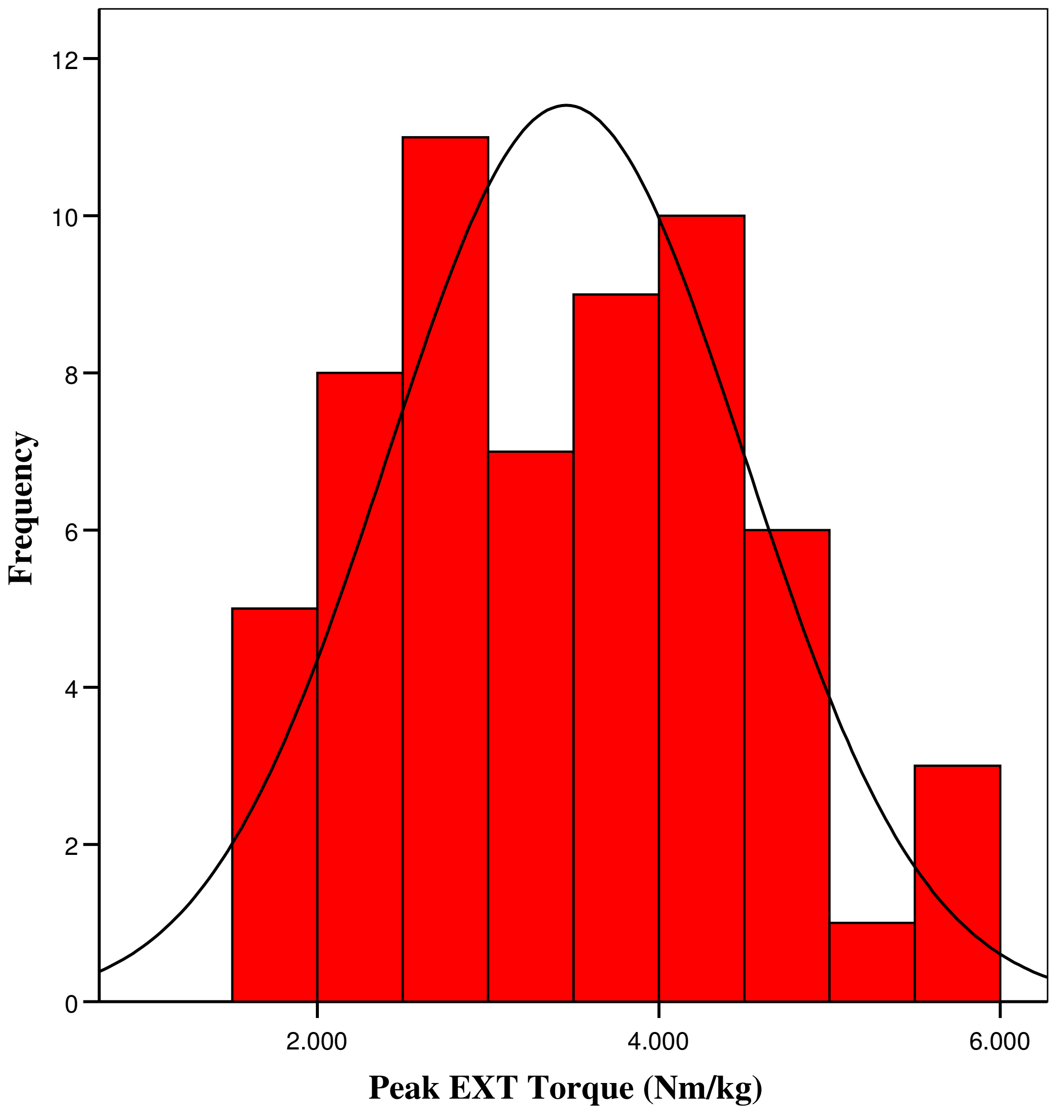
APPENDIX C_6. Navicular Drop Histogram



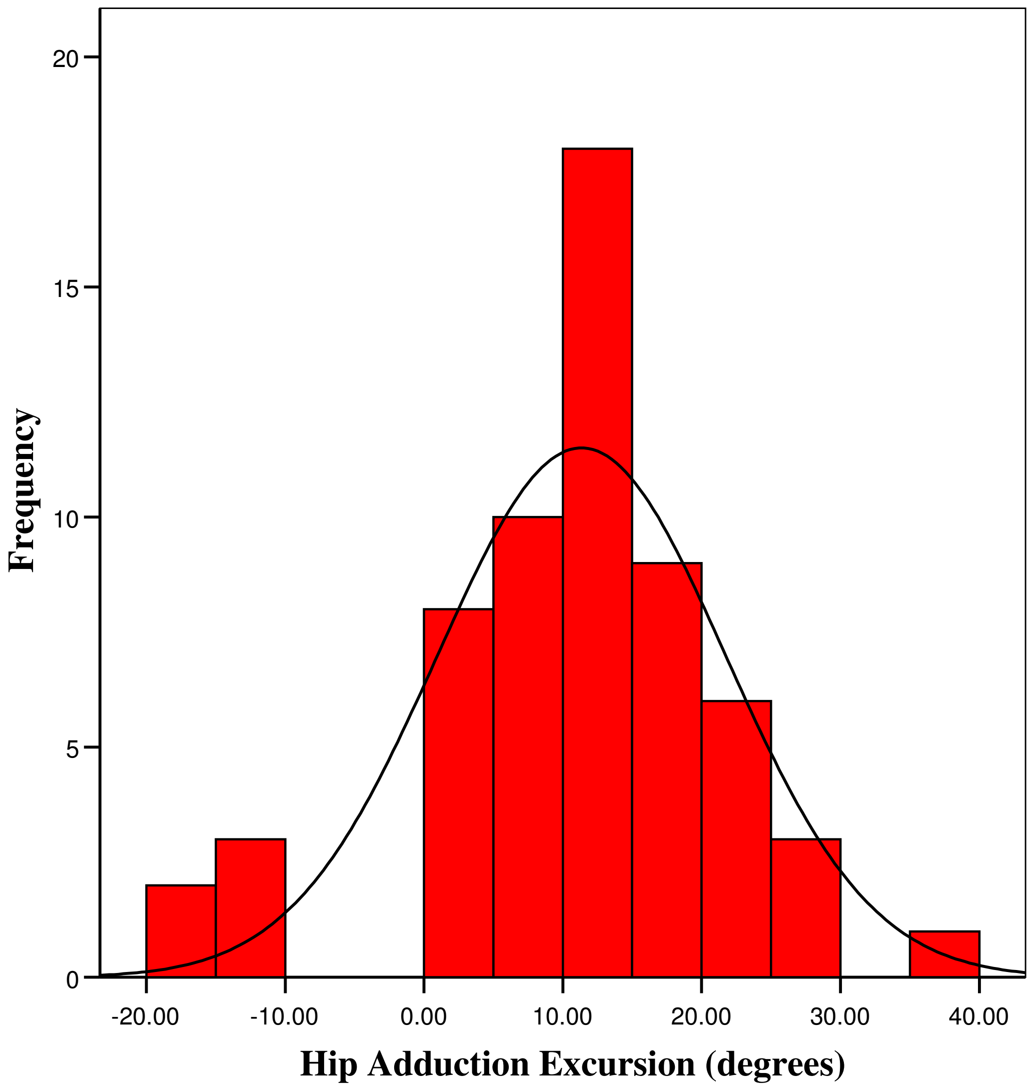
APPENDIX C_7. Hip Abduction Peak Torque Histogram



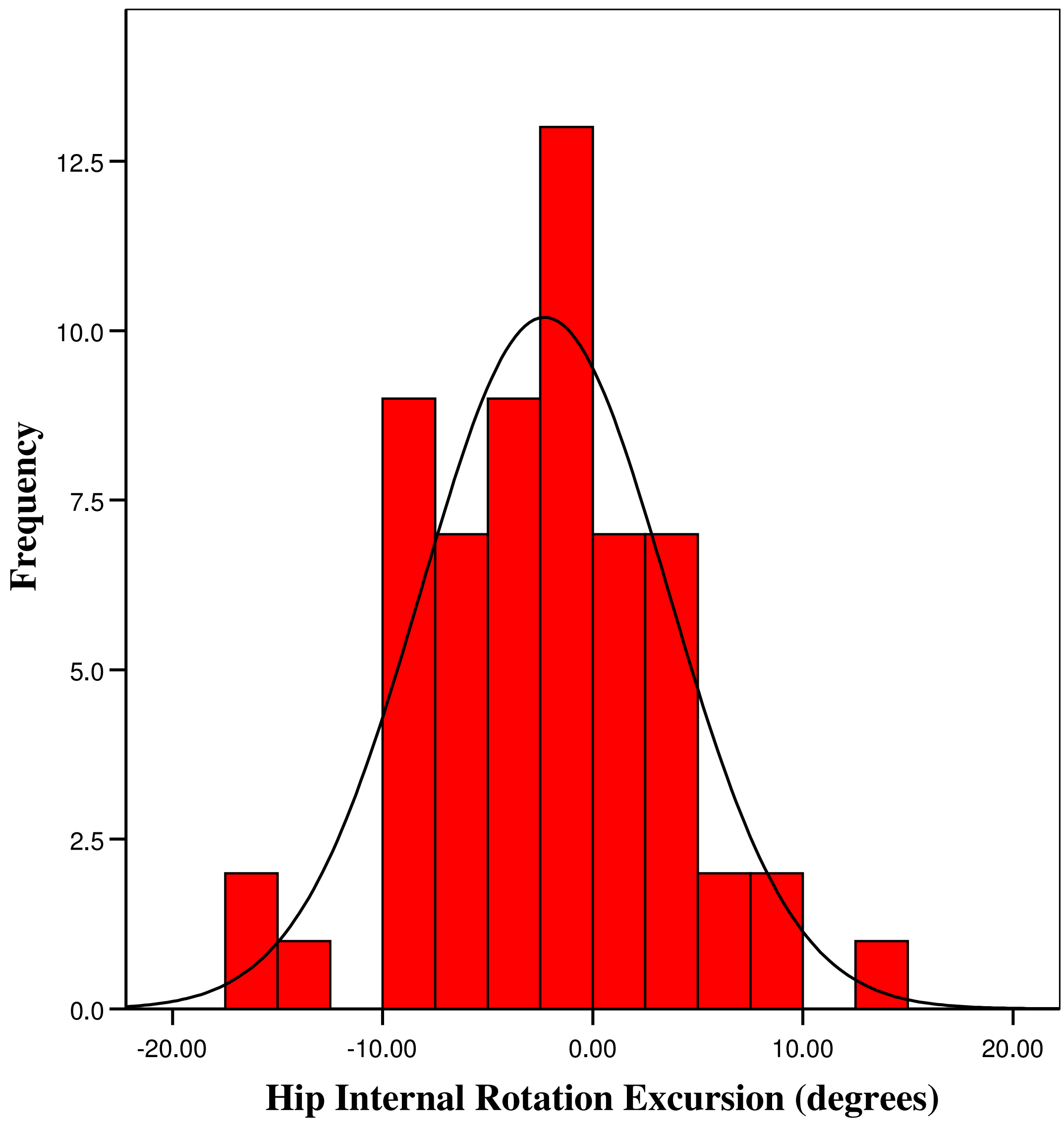
APPENDIX C_8. Hip Extension Peak Torque Histogram



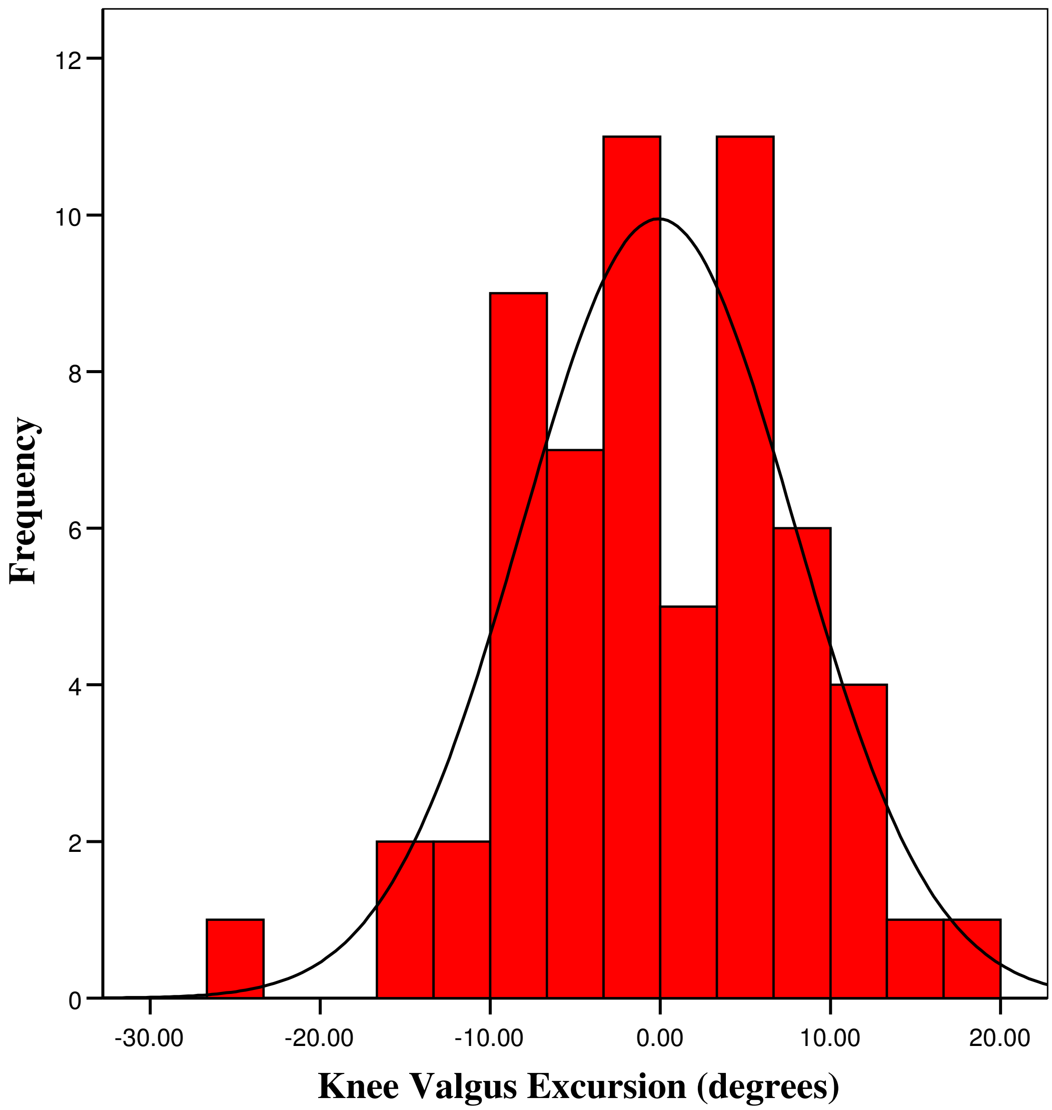
APPENDIX C_9. Hip Adduction Excursion Histogram



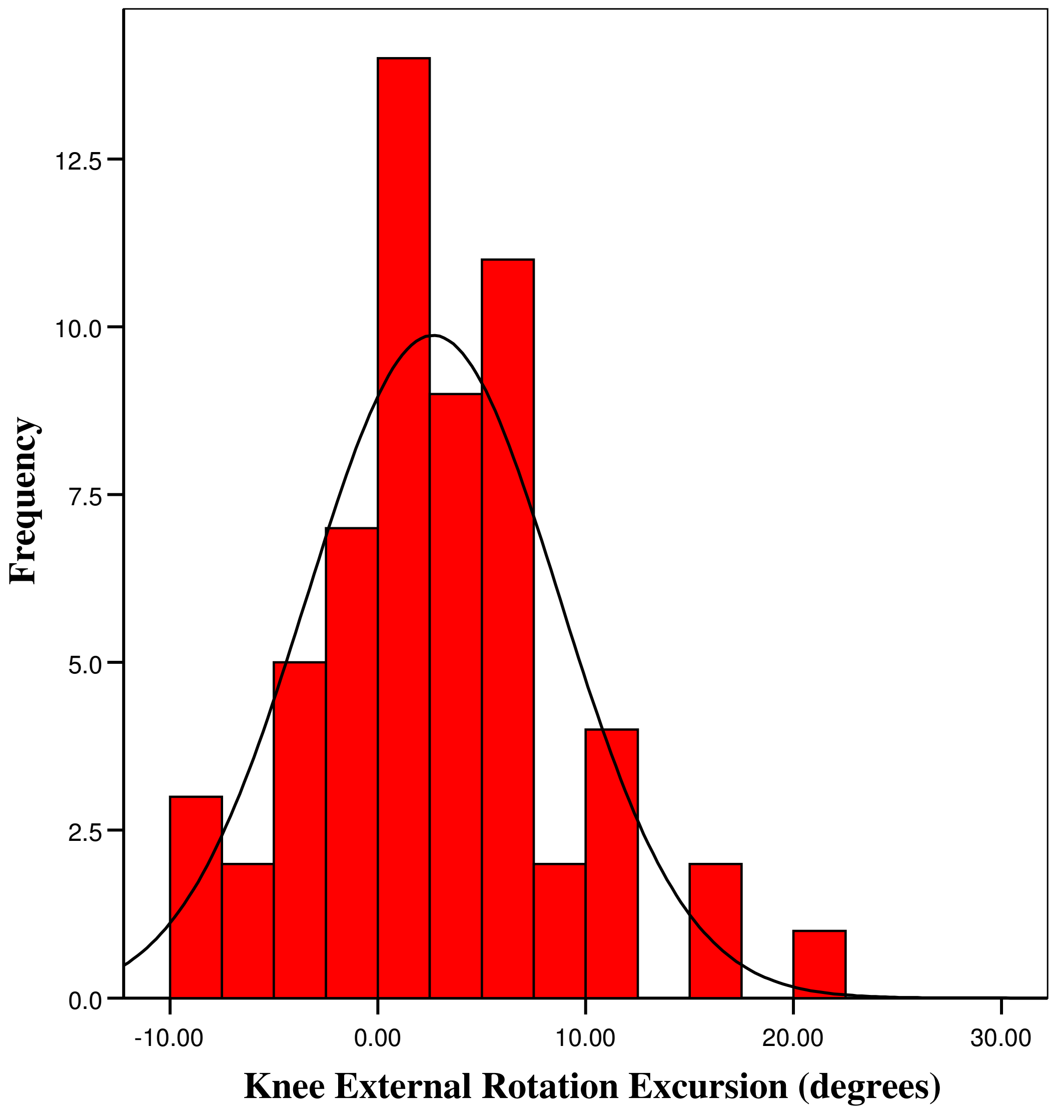
APPENDIX C_{10}. Hip Internal Rotation Excursion Histogram



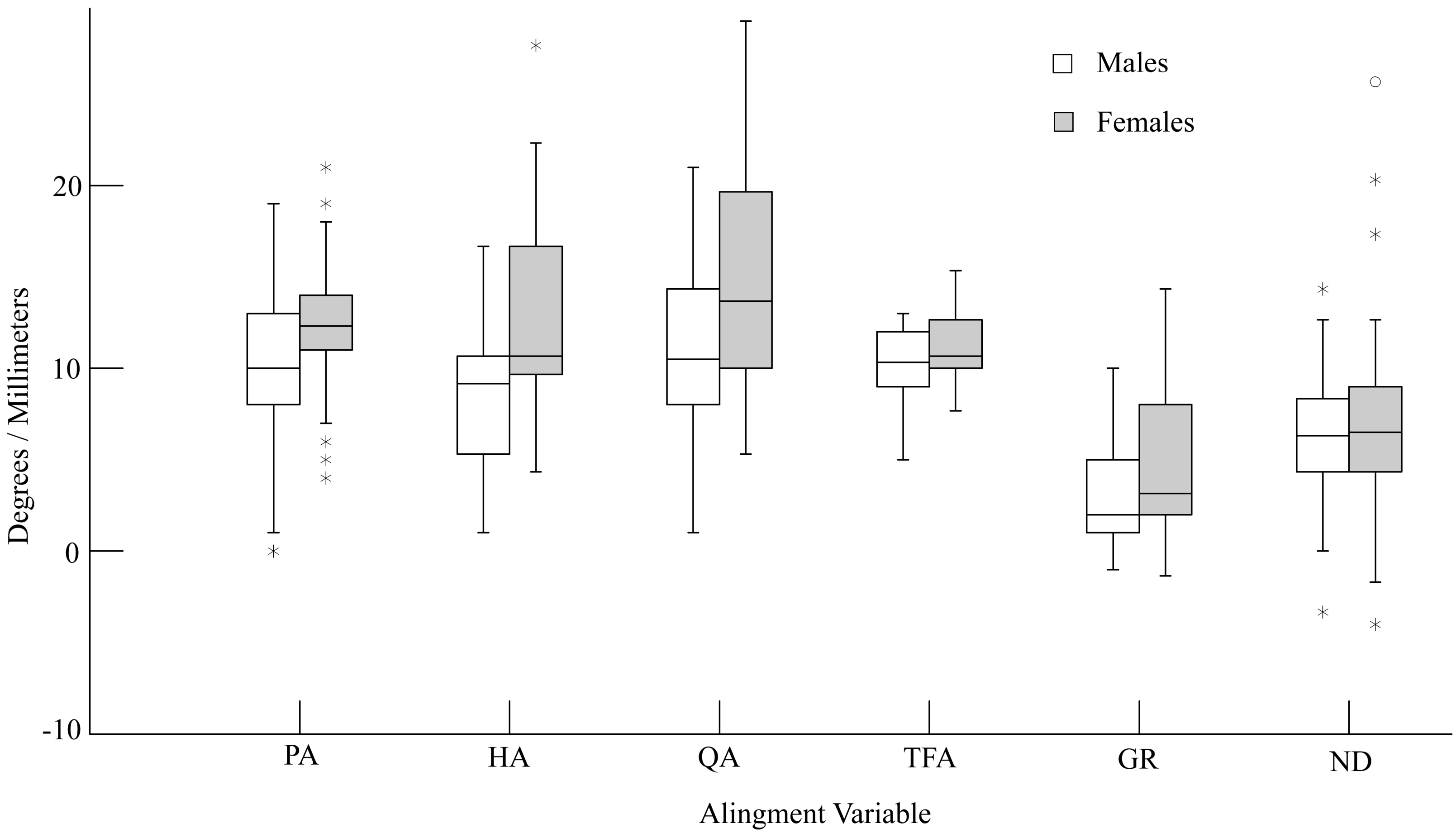
APPENDIX C_{11}. Knee Valgus Excursion Histogram



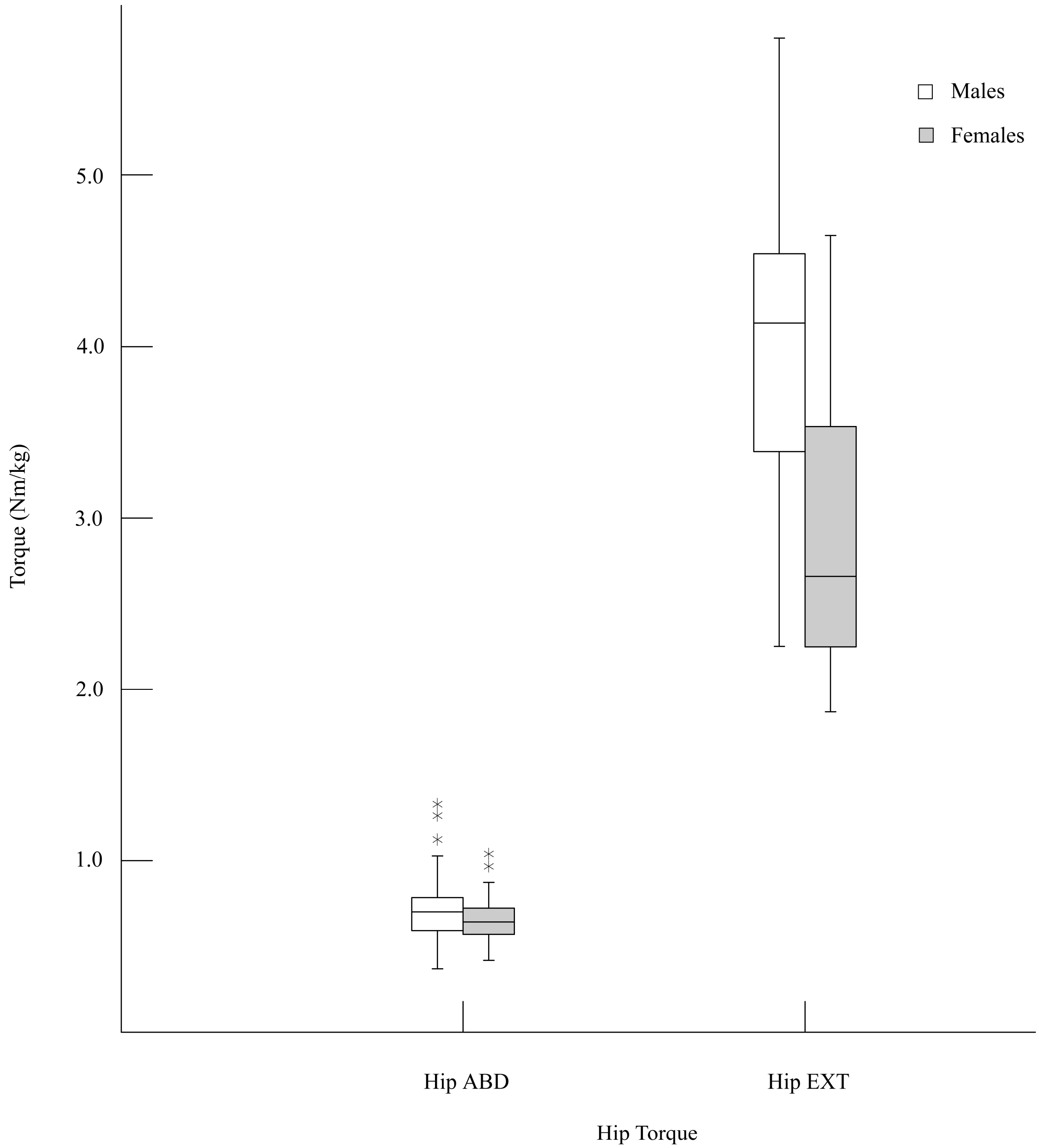
APPENDIX C_{12}. Knee External Rotation Excursion Histogram



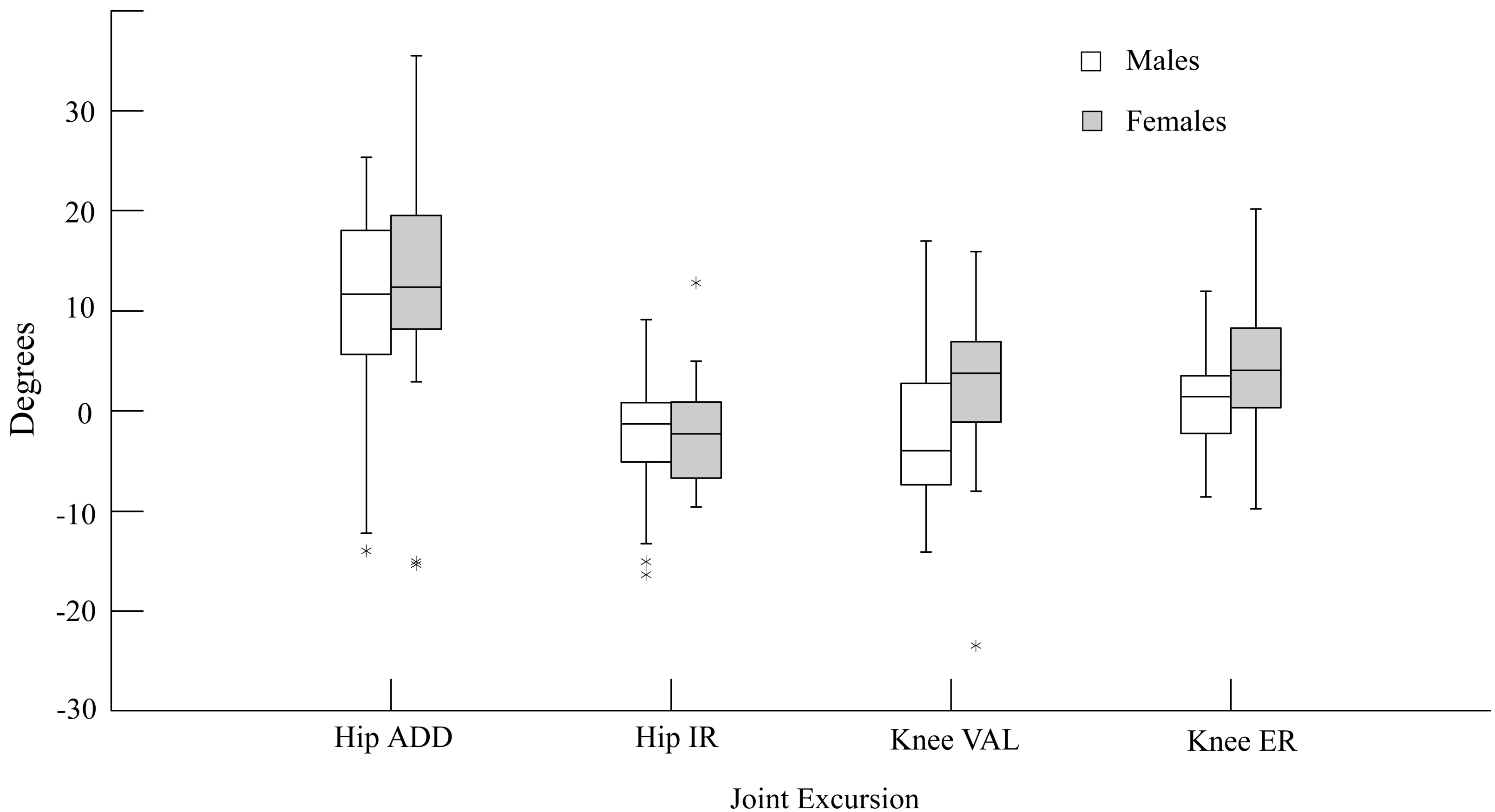
APPENDIX D_1. Lower Extremity Posture Box Plots by Sex



APPENDIX D_2. Hip Peak Torque Box Plots by Sex



APPENDIX D_3. Joint Excursion Box Plots by Sex



APPENDIX E. Subject Demographics, LEP and Hip Torque

Subject	Age (yrs)	Sex	Height (cm)	Mass (kg)	Dominant Limb	PA (°)	HA (°)	QA (°)	TFA (°)	GR (°)	ND (mm)	ABD torque (Nm/kg)	EXT torque (Nm/kg)
1	24	F	116.6	55.3	Left	11.0	14.0	16.3	10.7	2.0	7.3	0.45	4.52
2	20	F	149.8	57.5	Left	12.0	7.3	14.3	12.3	2.0	5.7	0.57	3.63
3	21	F	162.6	57.2	Left	16.0	13.7	13.7	9.7	3.0	9.0	0.72	2.36
4	22	F	154.9	85.2	Left	13.0	7.0	6.3	10.3	10.0	20.3	0.58	2.64
5	21	M	179.7	73.3	Left	1.0	7.3	11.0	11.0	6.7	6.7	0.46	4.16
6	20	F	158.0	46.4	Left	7.0	7.5	7.3	9.0	8.0	9.3	0.69	2.87
7	20	F	170.6	50.7	Left	12.0	12.3	12.0	12.7	3.0	6.7	0.65	2.39
8	20	F	170.0	62.2	Left	4.0	9.7	12.0	11.7	12.0	4.3	0.63	2.96
9	20	M	176.7	81.6	Left	9.7	5.3	15.3	11.0	3.0	4.7	0.64	2.80
10	29	F	169.6	69.7	Left	11.0	10.7	14.7	15.3	4.7	-1.7	0.86	3.23
11	20	M	164.8	87.9	Left	16.0	3.7	9.3	10.0	1.0	7.0	0.80	4.15
12	20	M	180.0	81.0	Left	19.0	6.7	15.3	9.3	5.7	14.3	0.67	2.58
13	25	F	157.2	53.2	Left	18.0	10.3	29.0	10.3	4.0	6.3	0.53	2.73
14	21	F	155.0	49.0	Left	19.0	9.0	23.3	13.3	13.3	25.7	0.78	3.53
15	22	M	181.9	100.5	Left	15.0	2.0	14.0	13.0	0.0	9.3	1.03	4.43
16	20	M	152.3	52.3	Left	12.0	4.3	15.3	13.0	2.0	7.7	0.59	3.20
17	20	F	156.4	62.6	Left	14.0	9.7	12.0	12.0	3.7	5.7	0.58	2.56
18	20	M	188.5	116.5	Left	10.0	2.3	7.3	12.0	1.7	0.0	0.86	4.29
19	21	F	157.0	54.6	Left	5.0	4.3	10.0	10.0	10.3	17.3	0.68	1.87
20	20	F	153.2	52.5	Left	11.0	9.3	14.0	10.0	14.3	11.7	0.57	2.57
21	25	M	185.5	103.2	Left	9.0	7.0	14.3	8.7	4.0	5.0	1.33	4.97
22	23	F	162.8	60.7	Left	8.3	10.3	9.7	10.7	0.7	8.3	1.04	1.98
23	26	F	164.5	66.4	Left	8.0	12.7	12.7	13.0	3.0	4.7	0.72	2.14
24	27	F	162.0	69.0	Left	11.0	10.0	19.7	9.7	0.3	4.3	0.45	1.93
25	23	M	187.4	110.9	Right	9.0	11.0	14.7	9.0	6.0	6.7	0.60	3.48

Subject	Age (yrs)	Sex	Height (cm)	Mass (kg)	Dominant Limb	PA (°)	HA (°)	QA (°)	TFA (°)	GR (°)	ND (mm)	ABD torque (Nm/kg)	EXT torque (Nm/kg)
26	20	M	152.4	49.1	Left	17.0	9.3	11.3	9.0	5.7	7.3	1.26	4.54
27	22	F	161.0	55.4	Left	14.0	16.7	10.0	10.7	0.0	6.3	0.67	2.24
28	21	M	171.1	88.4	Left	13.0	12.3	7.0	12.0	3.0	12.7	0.84	3.94
29	23	F	161.5	54.9	Left	21.0	19.3	21.7	8.0	0.3	4.3	0.42	2.68
30	31	M	171.6	89.3	Left	13.0	10.7	9.3	10.3	0.0	3.3	0.41	2.25
31	22	M	181.0	80.5	Left	3.0	10.0	12.7	7.0	5.0	5.3	0.71	5.80
32	23	M	167.6	78.0	Left	7.0	5.7	11.3	12.0	2.0	6.3	0.78	3.75
33	21	F	173.4	63.2	Left	4.0	14.3	12.0	14.0	10.0	-1.7	0.44	2.25
34	19	F	157.0	59.0	Right	14.0	10.7	15.3	12.0	7.3	8.0	0.97	4.04
35	21	F	159.1	58.8	Left	12.0	10.0	28.7	14.0	-1.3	6.3	0.63	3.05
36	21	F	168.2	73.0	Left	11.0	19.3	10.7	11.0	10.0	2.0	0.63	2.37
37	22	F	164.5	58.5	Left	15.0	19.0	15.0	9.0	3.3	7.3	0.77	4.23
38	21	F	166.2	61.0	Left	14.0	14.3	13.7	10.0	5.0	8.0	0.44	2.82
39	23	M	191.2	81.5	Left	5.0	16.7	9.7	9.0	3.0	1.0	0.77	3.39
40	24	M	180.4	113.5	Left	2.0	12.7	12.0	13.0	1.0	1.0	0.37	3.76
41	21	M	173.5	64.9	Left	15.0	7.0	7.3	11.7	3.3	6.3	0.54	3.11
42	20	F	169.4	57.1	Right	18.0	27.7	25.3	10.0	0.0	4.3	0.74	3.86
43	24	M	179.4	78.7	Right	9.0	15.3	8.0	5.0	2.0	8.3	0.64	4.12
44	25	M	181.1	89.1	Left	7.0	9.3	9.0	12.3	2.0	-3.3	0.79	4.24
45	29	M	191.9	85.7	Left	11.0	9.3	10.0	10.3	0.0	4.7	0.57	4.94
46	23	F	167.0	74.9	Left	12.7	19.3	10.0	15.0	-0.7	-4.0	0.65	2.28
47	31	M	179.7	77.0	Left	8.0	10.7	14.3	11.3	-1.0	10.3	0.75	3.86
48	21	M	182.3	84.5	Left	0.0	1.0	1.0	8.0	10.0	5.7	0.76	3.85
49	21	M	185.4	78.3	Left	9.0	11.3	4.0	12.0	5.0	5.0	0.61	5.47
50	23	F	156.5	59.5	Left	14.0	9.0	20.0	11.3	2.0	7.0	0.87	3.55
51	21	F	155.9	50.7	Left	16.0	19.7	20.7	10.0	3.0	12.7	0.47	1.95
52	22	F	169.6	65.5	Left	14.0	9.7	9.7	13.3	4.0	6.3	0.70	1.92
53	25	M	174.4	72.9	Left	14.0	9.7	10.0	11.0	-1.0	8.3	0.58	4.65

Subject	Age (yrs)	Sex	Height (cm)	Mass (kg)	Dominant Limb	PA (°)	HA (°)	QA (°)	TFA (°)	GR (°)	ND (mm)	ABD torque (Nm/kg)	EXT torque (Nm/kg)
54	25	M	179.7	82.0	Left	11.0	9.3	8.0	8.0	-0.7	3.0	0.70	5.68
55	30	M	190.3	92.3	Left	10.0	9.0	5.7	9.0	7.0	9.7	0.78	2.74
56	27	M	181.2	81.9	Left	11.0	6.3	7.3	8.0	8.0	8.3	1.12	5.60
57	27	M	177.2	70.1	Left	17.0	5.0	19.0	11.0	1.0	0.0	0.73	4.35
58	29	F	171.5	80.3	Left	6.0	22.3	5.3	7.7	3.0	9.0	0.64	4.65
59	30	M	169.5	67.9	Left	11.0	5.3	13.7	8.0	2.0	8.0	0.62	4.27
60	25	M	183.9	76.1	Right	8.0	14.3	21.0	10.0	0.7	4.3	0.57	3.23

APPENDIX F. Raw Data for Joint Excursions During Single Leg Squat

Subject	Hip ADD Excursion (deg)	Hip IR Excursion (deg)	Knee VAL Excursion (deg)	Knee ER Excursion (deg)
1	8.19	-3.62	2.42	5.97
2	3.38	-0.57	-4.02	0.97
3	23.88	-4.98	6.90	-4.56
4	14.56	4.76	-8.03	9.38
5	19.23	8.47	-9.69	5.56
6	28.40	-5.88	15.97	0.81
7	11.68	-1.43	4.68	6.89
8	8.22	-2.79	0.12	11.45
9	20.46	-15.04	7.24	11.98
10	23.84	-9.56	5.21	8.29
11	12.04	-3.68	-0.49	1.13
12	16.28	-4.56	11.97	5.62
13	12.87	3.45	3.95	0.30
14	17.16	-5.15	10.60	7.33
15	12.28	-3.06	-5.27	4.88
16	18.05	7.18	-4.55	0.77
17	13.54	-9.12	-1.12	0.45
18	3.27	-13.26	-2.21	-4.05
19	2.93	4.99	-7.19	-3.18
20	19.53	-1.81	0.89	4.57
21	19.76	-9.82	9.00	3.53
22	4.96	0.87	-1.06	-2.21
23	9.59	-9.41	8.46	6.77
24	13.08	-6.88	6.36	-1.58
25	-11.99	9.15	-7.32	0.71
26	18.41	-8.87	4.18	-4.07
27	10.16	-8.90	12.72	6.19
28	11.31	2.45	2.78	1.68
29	12.76	-9.33	3.29	0.31
30	25.36	-16.35	17.01	3.17
31	8.17	5.85	-7.23	5.35
32	5.64	-0.27	-10.60	-0.73
33	29.12	-1.26	3.83	20.17
34	-15.06	-1.27	-0.30	-9.76
35	7.17	-5.80	3.73	2.55
36	13.39	-0.84	-3.18	6.32
37	23.54	0.07	8.92	11.68
38	11.99	-6.73	9.92	15.35
39	13.87	-5.60	4.46	7.14

Subject	Hip ADD Excursion (deg)	Hip IR Excursion (deg)	Knee VAL Excursion (deg)	Knee ER Excursion (deg)
40	20.42	-1.69	-1.58	6.82
41	4.28	-0.19	-2.58	2.56
42	-15.34	12.82	-23.47	-9.73
43	-13.98	-1.03	-9.08	1.06
44	14.47	-3.34	-4.16	-0.48
45	2.93	-5.14	-7.39	-5.77
46	4.65	-9.56	-5.44	1.23
47	16.40	0.62	-0.03	0.78
48	9.80	-1.58	-7.38	-2.35
49	8.01	3.77	-9.82	2.11
50	19.80	3.20	-2.94	-1.02
51	10.06	-3.96	6.27	11.13
52	11.56	2.14	10.06	3.56
53	22.21	-0.06	4.56	2.32
54	8.97	-0.42	-3.78	-2.25
55	2.95	-7.81	-14.09	-5.86
56	6.33	0.00	-13.79	-8.60
57	10.96	-4.26	-1.91	2.74
58	35.54	4.05	6.56	15.58
59	12.46	2.67	-11.74	-4.51
60	-12.21	0.81	-6.60	3.17

APPENDIX G. Raw Data for sEMG of the Posterio-Lateral Hip

Subject	Gmed MVIC Peak Amp (volts)	Gmax MVIC Peak Amp (volts)	Gmed Initial 20% of SLS (% MVIC)	Gmed Final 20% of SLS (% MVIC)	Gmed Total of SLS (% MVIC)	Gmax Initial20% of SLS (% MVIC)	Gmax Final20% of SLS (% MVIC)	Gmax Total of SLS (% MVIC)
1	0.501	0.525	0.131	0.233	0.184	0.041	0.046	0.042
2	0.225	0.337	0.167	0.169	0.168	0.076	0.067	0.071
3	0.462	0.156	0.162	0.215	0.192	0.620	0.377	0.463
4	0.270	0.080	0.323	0.456	0.405	0.273	0.246	0.258
5	0.206	0.194	0.584	0.797	0.717	0.086	0.125	0.101
6	1.241	0.199	0.108	0.227	0.171	0.183	0.249	0.206
7	0.413	0.579	0.294	0.529	0.426	0.053	0.043	0.047
8	0.491	0.172	0.131	0.345	0.233	0.093	0.143	0.116
9	0.551	0.067	0.164	0.292	0.219	0.406	0.332	0.346
10	0.613	0.324	0.299	0.422	0.392	0.103	0.145	0.199
11	0.246	0.065	0.166	0.233	0.214	0.160	0.159	0.138
12	0.306	0.069	0.293	0.502	0.377	0.241	0.222	0.214
13	0.281	0.118	0.231	0.387	0.328	0.288	0.241	0.248
14	0.305	0.137	0.297	0.497	0.447	0.118	0.137	0.137
15	0.343	0.173	0.286	0.511	0.390	0.084	0.164	0.123
16	0.486	0.213	0.160	0.211	0.175	0.138	0.226	0.170
17	0.245	0.043	0.142	0.220	0.172	0.153	0.153	0.150
18	0.490	0.097	0.107	0.325	0.210	0.123	0.147	0.134
19	0.609	0.069	0.107	0.111	0.107	0.218	0.219	0.212
20	0.854	0.305	0.083	0.130	0.110	0.057	0.167	0.108
21	0.682	0.506	0.082	0.166	0.117	0.053	0.115	0.080
22	0.187	0.018	0.235	0.226	0.239	1.137	0.999	1.043
23	0.245	0.111	0.127	0.241	0.163	0.168	0.299	0.236

Subject	Gmed MVIC Peak Amp (volts)	Gmax MVIC Peak Amp (volts)	Gmed Initial 20% of SLS (% MVIC)	Gmed Final 20% of SLS (% MVIC)	Gmed Total of SLS (% MVIC)	Gmax Initial20% of SLS (% MVIC)	Gmax Final20% of SLS (% MVIC)	Gmax Total of SLS (% MVIC)
24	0.271	0.026	0.183	0.278	0.238	0.512	0.609	0.543
25	0.175	0.091	0.374	0.405	0.386	0.173	0.225	0.196
26	0.643	0.202	0.102	0.110	0.105	0.085	0.215	0.138
27	0.387	0.057	0.151	0.308	0.205	0.658	0.845	0.716
28	0.221	0.124	0.167	0.418	0.330	0.125	0.122	0.127
29	0.820	0.144	0.320	0.390	0.345	0.489	0.615	0.521
30	0.104	0.030	0.509	0.538	0.532	0.963	0.525	0.656
31	0.840	0.374	0.110	0.212	0.171	0.053	0.094	0.080
32	0.389	0.157	0.148	0.338	0.234	0.099	0.160	0.118
33	0.418	0.275	0.235	0.495	0.353	0.079	0.170	0.125
34	0.300	0.163	0.187	0.178	0.203	0.114	0.148	0.131
35	0.213	0.108	0.153	0.177	0.166	0.138	0.160	0.153
36	0.170	0.085	0.421	0.577	0.502	0.465	0.347	0.385
37	0.387	0.188	0.311	0.344	0.359	0.146	0.157	0.151
38	0.168	0.108	0.443	0.542	0.531	0.186	0.259	0.244
39	0.541	0.190	0.129	0.263	0.204	0.072	0.125	0.097
40	0.207	0.062	0.224	0.222	0.221	0.218	0.379	0.295
41	0.567	0.243	0.211	0.363	0.284	0.076	0.107	0.089
42	0.589	0.223	0.116	0.167	0.146	0.102	0.166	0.123
43	0.674	0.372	0.063	0.227	0.150	0.047	0.051	0.050
44	0.513	0.149	0.126	0.194	0.162	0.137	0.186	0.159
45	0.314	0.499	0.259	0.441	0.407	0.033	0.066	0.045
46	0.253	0.135	0.150	0.405	0.272	0.297	0.447	0.377
47	0.217	0.416	0.209	0.359	0.281	0.029	0.041	0.036
48	0.348	0.258	0.183	0.394	0.331	0.068	0.065	0.064
49	0.342	0.504	0.181	0.264	0.256	0.021	0.029	0.025
50	0.482	0.203	0.142	0.355	0.294	0.079	0.181	0.131

Subject	Gmed MVIC Peak Amp (volts)	Gmax MVIC Peak Amp (volts)	Gmed Initial 20% of SLS (% MVIC)	Gmed Final 20% of SLS (% MVIC)	Gmed Total of SLS (% MVIC)	Gmax Initial20% of SLS (% MVIC)	Gmax Final20% of SLS (% MVIC)	Gmax Total of SLS (% MVIC)
51	0.522	0.069	0.156	0.268	0.212	0.389	0.401	0.375
52	0.474	0.072	0.164	0.247	0.206	0.158	0.202	0.186
53	0.549	0.298	0.120	0.163	0.128	0.055	0.094	0.087
54	0.480	1.045	0.146	0.281	0.211	0.040	0.043	0.045
55	0.240	0.128	0.251	0.803	0.590	0.274	0.270	0.253
56	0.674	0.372	0.098	0.186	0.146	0.028	0.043	0.034
57	0.750	0.594	0.200	0.319	0.274	0.073	0.109	0.101
58	0.235	0.541	0.186	0.243	0.227	0.082	0.044	0.077
59	0.892	0.739	0.210	0.348	0.311	0.020	0.030	0.026
60	0.532	0.116	0.229	0.318	0.265	0.085	0.112	0.095