HOGG, JENNIFER A., Ph.D. The Influence of Femoral Structure, Hip Capsular Constraints, and Gluteal Muscle Strength and Activation on Temporal Patterns of Functional Valgus Collapse. (2018)
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Functional valgus collapse (a combination of knee abduction and internal rotation and hip adduction and internal rotation) is a modifiable lower extremity movement pattern commonly associated with anterior cruciate ligament (ACL) injuries in females. Though the gluteus maximus and gluteus medius have frequently been named contributors to functional valgus collapse, evidence supporting their role in lower extremity movement has been inconsistent, and could in part be due to methodological differences between studies and the accepted practice of analyzing discrete variables instead of overall movement patterns. Better elucidation of gluteal muscle influence on lower extremity biomechanics may be a critical step for the reduction of ACL injury rates, as neuromuscular dysfunction is likely more responsive to injury prevention efforts than are other risk factors such as bony anatomy, ligament quality, or hormonal influences, that are more difficult to modify. Therefore, the purpose of this study was to 1) describe the neuromechanical profiles throughout the landing phase of single-leg and double-leg forward landings in males and females, 2) quantify the contributions of gluteal muscle strength and activation to peak angles and moments of functional valgus collapse after controlling for one's femoral alignment, and 3) explore the association between gluteal muscle function and overall functional valgus collapse throughout the landing phase.

To accomplish this, 45 females and 45 males with no history of knee surgery were measured for femoral anteversion, hip ROM, and hip strength and then underwent biomechanical testing during single-leg and double-leg forward landings to examine muscle activation and 3-dimensional biomechanics. Data were analyzed using conventional group and correlative analyses and also with statistical parametric mapping (SPM), which allowed for a more

comprehensive examination of the entire biomechanical time series. Biomechanical variables of interest included joint angles and moments comprising functional valgus collapse: hip adduction and internal rotation and knee abduction and internal rotation.

In the comparison between single-leg and double-leg landings by sex, sex differences in the frontal plane were task dependent, though females maintained greater absolute knee abduction and hip adduction throughout the landing phases. Sex by task interactions revealed that females landed with smaller knee adduction angles than males, particularly during the single-leg landing (p=.03), while females' knee abduction excursion was greater than males', particularly during the double-leg landing (p=.01). Across task, females displayed 4.1° greater peak knee abduction than males (p=.002), and this was specific to 37-46% of the landing phase (p=.05). Females went through 1.0° more hip abduction than males (p=.05), and used a smaller proportion of their gluteus maximus (p=.01) in both tasks.

Examination of gluteal muscle contribution to individual and overall levels of functional valgus collapse in females revealed that at the 18% and 20% time points during the landing phase, less hip abduction strength and greater gluteus medius activation predicted greater peak hip adduction angles (R^2 change = .10; p = .02) and higher external hip adduction moments (R^2 change = .14, p = .06). Greater hip extension strength predicted greater peak hip abduction angles (R^2 change = .08; p = .05), while greater gluteus maximus activation strengthened the prediction of greater initial (R^2 change = .10, p = .03) and peak (R^2 change = .14, p = .01) knee internal rotation angles. From 7% - 8% of the landing phase, greater external rotation ROM was associated with greater external hip adduction moment (R^2 change = .18, p = .01).

In males, less hip abduction strength strengthened the prediction of greater initial (R^2 change = .12, p = .01) and peak knee internal rotation angles (R^2 change = .14, p = .01), lesser peak knee external rotation angles (R^2 change = .07, p =.09), and lesser peak knee abduction moments (R^2 change = .06, p =.11). Less hip extension strength with greater gluteus maximus activation predicted greater peak hip external rotation moments (R^2 change = .14, p = .01). Specifically from the 3% - 9% time points of the landing phase, greater hip extension strength was associated with greater knee abduction moment (R^2 change = .17, p = .01) and less hip adduction moment (R^2 change = .24, p = .001). At 0% and from 2% - 3% of the landing phase, greater internal and external rotation ROM were associated with greater knee abduction angle (R^2 change = .27, p = .01) and greater hip adduction angle (R^2 change = .23, p = .02).

These results indicate that lower extremity biomechanics during a single-leg landing task are appreciably different than those observed during a double-leg landing task, and that a single-leg landing task elicits more profound sex differences, particularly during the early stage of single-leg load acceptance when ACL injuries are thought to occur (30-40ms post initial ground contact). As such, a single-leg landing task may be more appropriate for biomechanical screening of ACL injury risk. Gluteal strength and activation explained a unique proportion of variance in lower extremity biomechanics beyond what was explained by femoral alignment. In females, weaker gluteal muscles predicted riskier frontal plane hip kinematics. In males, gluteal function was more associated with kinetics. This implies that our male cohort used their musculature to create torque about a joint, whereas our female cohort was unable to create torque. Though femoral alignment (total ROM) explained considerably greater proportions of biomechanical variance than did gluteal function, observed associations between gluteal muscle function and biomechanics occurred 10-20ms after associations between femoral alignment and biomechanics. While the gluteal muscles may act mechanically independent of femoral alignment, it is possible that gluteal muscle function could be temporally linked to one's femoral alignment. With these findings in mind, it may be beneficial for clinicians to implement gluteal strengthening programs and to encourage gluteal muscle pre-activation in individuals with excessive hip ROM to lessen their propensity for functional valgus collapse.

THE INFLUENCE OF FEMORAL STRUCTURE, HIP CAPSULAR CONSTRAINTS, AND GLUTEAL MUSCLE STRENGTH AND ACTIVATION ON TEMPORAL PATTERNS OF FUNCTIONAL VALGUS COLLAPSE

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APPROVAL PAGE

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CHAPTER I

INTRODUCTION

Of the more than 350,000 anterior cruciate ligament (ACL) injuries that occur annually in the United States, an estimated 72% occur through non-contact mechanisms (Moses, Orchard, & Orchard, 2012a; Wojtys & Brower, 2010). It is theorized that functional valgus collapse, a non-contact mechanism comprised of knee abduction, tibial internal rotation, hip adduction, and hip internal rotation, may increase the potential for ACL injury (Hewett et al., 2005; Ireland, 1999). Retrospective videographic studies have consistently reported the presence of a valgus knee collapse during ACL injury, particularly in females, as evidenced by increased pronation, increased medial knee collapse, increased hip adduction, and greater ipsilateral trunk lean (Boden, Torg, Knowles, & Hewett, 2009; Krosshaug et al., 2007). In vitro research has corroborated the injurious nature of functional valgus collapse suggested by videographic evidence. Specifically, the combination of internal tibial rotation and anterior tibial translation increased ACL strain greater than either internal tibial rotation or anterior tibial translation alone (Berns, Hull, & Patterson, 1992; Fukuda et al., 2003; Kiapour et al., 2014; Tron Krosshaug et al., 2007; Markolf et al., 1995). The strain resulting from combined internal tibial rotation and anterior tibial translation was further increased by the addition of a pure frontal plane valgus force (Berns et al., 1992). Of note, in the absence of tibial rotation and anterior tibial translation, a pure valgus force only minimally increased ACL strain, if at all (Berns et al., 1992; Markolf et al., 1995; Y. Oh, Ashton-Miller, & Wojtys, 2011). Conversely, an isolated tibial internal rotation torque, of a magnitude common in athletics, was sufficient to rupture the ACL (Meyer & Haut, 2008). This agreed well with research reporting the ACL to have greater sensitivity to rotational moments than to frontal plane moments (Y. K. Oh et al., 2012). Together, these studies indicate that loads causing ligament rupture likely have a rotational, transverse plane component in addition to frontal plane movements, suggesting that ACL injuries may result from multiplanar loading patterns.

It is accepted that lower extremity movement acts occurs in a kinematics chain fashion, such that anterior pelvic tilt is thought to pair with greater femoral internal rotation, internal tibial rotation, and pronation (Duval, Lam, & Sanderson, 2010; Khamis & Yizhar, 2007). It is also accepted that the knee, hip, and trunk are mechanically coupled via ground reaction forces (Hewett & Myer, 2011; Imwalle et al., 2009). Given a ground reaction force that passes lateral to the knee joint, an adducted hip and an ipsilateral trunk lean become necessary to maintain an upright posture (Timothy E Hewett & Myer, 2011; Sigward & Powers, 2007a). Empirical evidence has demonstrated this coupling, showing that hip adduction alone may account for as much as 25% of the variance in knee abduction during cutting maneuvers (Imwalle, Myer, Ford, & Hewett, 2009). Along with greater vertical GRF and increased hip adduction, increased hip internal rotation has also been shown to contribute to increased knee valgus angles and moments during cutting maneuvers (R^2=.36-.62) (Havens & Sigward, 2014; Sigward & Powers, 2007a). Taking into account an integrated movement strategy and the evidentiary transverse and frontal plane coupling of these joints, controlling adduction and internal rotation of the hip may be an imperative step in the prevention of functional valgus collapse.

Femoral anteversion and passive hip range of motion (ROM) are two anatomical hip characteristics thought to influence dynamic hip adduction and internal rotation (Howard et al., 2011; Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011), and thus functional valgus collapse. Specifically, increased femoral anteversion and greater internal rotation hip ROM are suggested to bias the femur toward internal rotation and adduction across various functional tasks, thus predisposing one towards greater knee valgus (A.-D. Nguyen et al., 2015; A Nguyen, Cone, Stevens, Schmitz, & Shultz, 2009; Sigward, Ota, & Powers, 2008). Because females are known to have greater amounts of both femoral anteversion and hip internal rotation ROM (Fan, Copple, Tritsch, & Shultz, 2014b; Moreno-Pérez, Ayala, Fernandez-Fernandez, & Vera-Garcia, 2015; A.-D. Nguyen & Shultz, 2007), this may in part account for the valgus collapse mechanism more commonly observed in females (T E Hewett, Torg, & Boden, 2009; Tron Krosshaug et al., 2007).

As the muscles primarily responsible for hip abduction and external rotation, the gluteus medius and gluteus maximus are often considered active restraints to dynamic hip adduction and internal rotation, respectively. As such, they have the potential to mediate the effects of hip range of motion and femoral anteversion. Despite copious literature investigating the gluteal muscles' contribution to valgus collapse, the evidence is mixed. Varying methodology between studies makes it difficult to compare findings. Both absolute torque generating capacity and electromyographic (EMG) muscle activation amplitude (as a % of maximal voluntary isometric contraction; MVIC) during functional tasks have been examined for their influence on dynamic hip adduction and internal rotation. At best, greater isometric hip abductor peak torque generation is moderately correlated with less hip adduction and knee valgus excursion (r= -.40 and -.35, respectively) (Jacobs et al., 2007). However, other similarly conducted studies found no significant relationships (Homan et al., 2013; Sigward et al., 2008; Thijs et al., 2007). External rotation isometric strength alone also yields mixed results (Cashman, 2012; Cronstrom, Creaby, Nae, & Ageberg, 2016; Howard, Fazio, Mattacola, et al., 2011). However, when muscle activation is included as a predictor, a more complete picture is rendered. Individuals with weaker hip abductors and external rotators have been shown to use greater percentages of their MVIC to complete a functional task (Homan et al., 2013a). This may explain why another study observed higher gluteal activation amplitude (% of MVIC) in those with greater knee valgus excursion during a single-leg squat (A.-D. Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011). Therefore, both hip muscle strength and activation may need to be accounted for when analyzing the influence of the gluteal muscles on functional valgus collapse.

While hip internal rotation and adduction appear to be critical components of functional valgus collapse, the combined impact of gluteal strength and activation and femoral anteversion and passive hip ROM has yet to be examined with regard to stabilizing the hip during sport activity. Examining these in combination is important, as the gluteal muscles may have the ability to mitigate potentially negative effects of high internal rotation ROM or femoral anteversion. Therefore, not only is it important to include passive hip ROM and femoral anteversion as predictor variables, but including MVIC values along with muscle activation amplitude may be necessary.

Statement of the Problem

Existing ACL injury prevention programs are designed to improve dynamic lower extremity alignment, and have been successful in reducing ACL injury risk (Taylor, Waxman, Richter, & Shultz, 2015). However, overall rates of ACL injury have remained constant over the past two decades (Arendt & Dick, 1995; Moses, Orchard, & Orchard, 2012b). This suggests that safer dynamic alignment is not being retained after completing an ACL injury prevention program. This could be the result of underlying structural characteristics, which are not modified by prevention programs. It is also possible that ACL injury prevention programs are targeting the wrong constructs. Because of this, it may be important to account for differences in structural alignment when examining influences of muscle activation on lower extremity biomechanics.

While femoral anteversion, passive hip ROM, and gluteal strength and activation in isolation have the potential to influence hip and knee control, the interaction of these factors to influence movement during a dynamic task has not yet been elucidated. While the existing evidence is inconclusive regarding gluteal influences on functional valgus collapse, previous studies have not analyzed muscle strength and activation in conjunction with femoral anteversion and passive hip ROM, nor have they used single-leg functional tasks to examine these relationships. Because demands on the lumbo-pelvic-hip complex are greater in a single-leg stance, using a single-leg task may better highlight gluteal contributions to functional valgus collapse. Furthermore, because females are more likely to display functional valgus collapse (T Krosshaug, Slauterbeck, Engebretsen, & Bahr, 2007), sex-specific research designs may be necessary to detect mechanistic patterns. Accounting for differences in transverse femoral alignment and capsular constraints within a sex-specific design may serve to better highlight gluteal impact on functional valgus collapse, and thus provide an avenue to affect biomechanical change in ongoing ACL injury prevention efforts.

Perhaps another reason for the inconclusive findings regarding gluteal influence on functional valgus collapse is that statistical approaches commonly used to analyze these data are limited. Functional valgus collapse is a coupled movement, exhibiting patterns unfolding over the course of a landing or cutting maneuver. Common practice is to collapse this movement pattern to a handful of discrete variables for analysis (e.g. initial contact, peak and excursion values), with each variable representing a single instant in time. Such analyses assume that movement occurs linearly, failing to take into account the possibility that prolonged joint loading or erratic movement may hold importance for ACL injury risk. Few studies have taken the full temporal nature of valgus collapse into account. Those that have were able to better identify loading and timing differences between participants with varying lower extremity alignment and laxity profiles. (A.-D. Nguyen et al., 2015; S. J. Shultz & Schmitz, 2009a). As such, employing a more holistic statistical technique may help to better characterize the impact of hip structure and gluteal muscle function on functional valgus collapse patterns. Understanding these factors and their influences on knee joint loading rates is critical to identifying potential modifiable risk factors to target in ACL injury prevention programs.

Objective and Hypotheses

The objective of this study was to determine the extent to which femoral anteversion, passive hip ROM, and gluteal strength and activation impact patterns of functional valgus collapse during a single-leg forward landing task in separate female and male cohorts.

Aim 1: Examine sex-specific biomechanics throughout the entire landing phases of single-leg and double-leg forward landing tasks.

Hypothesis 1a: Compared to males, females will exhibit greater functional valgus collapse, as exhibited by greater joint angles and external moments associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation. This pattern will be more pronounced in a single-leg forward landing than in a double-leg forward landing.

Hypothesis 1b: Statistical Parametric Mapping 2x2 ANOVAs, which examine biomechanical differences across the entire landing phase, will identify specific time points at which lower extremity biomechanics differ by task, and by sex, thus providing a more complete analysis than using discrete, singular time point variables.

Aim 2: Determine the extent to which femoral anteversion and passive internal and external rotation hip ROM are associated with functional valgus collapse during a single-leg forward landing task in females and males, and the extent to which these influences are mediated by gluteal muscle strength and activation, and whether these relationships become stronger and more specific once taking into account the timing and temporal nature of functional valgus collapse.

Hypothesis 2a: Greater femoral anteversion and greater internal rotation ROM and lesser external rotation ROM will predict greater movement toward functional valgus collapse during a single-leg forward landing task, as evidenced by increased joint angles and external moments associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation.

Hypothesis 2b: The relationship between increased femoral anteversion, increased hip internal rotation ROM, decreased external rotation ROM and components of functional valgus collapse (as evidenced by increased joint angles and external moments associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation) will be weaker once controlling for the mediating effect of gluteus maximus and gluteus medius strength and activation.

Hypothesis 2c: A statistical parametric mapping canonical correlation analysis, which takes into account the temporal nature of functional valgus collapse, will identify stronger relationships between hip structure and function with functional valgus collapse than will using conventional correlative analyses with discrete, singular time point variables.

Limitations and Assumptions

1. Findings from this dissertation are neither generalizable to populations other than young healthy females and males, nor to tasks other than the single-leg or double-leg forward landing.

2. Three-dimensional motion capture, as represented by The Phase Space IMPULSE motion tracking system, is a valid and reliable tool for measuring biomechanical kinematics.

3. Embedded forceplates, as represented by dual Bertec plates, are valid and reliable tools for capturing biomechanical kinetics.

4. Inverse dynamics is an adequate method of computing three dimensional joint forces.

5. Femoral anteversion, as measured by an inclinometer, is a suitable surrogate for radiographic measurement of femoral anteversion.

6. Passive hip ROM, as measured prone with an inclinometer, is representative of capsular restraints of the femoral head.

7. All participants gave a maximal effort during maximal voluntary isometric contraction (MVIC) strength testing.

8. Surface electromyographic amplitude is not analogous to force.

9. Surface electromyography is a valid and reliable method of measuring muscle activity during functional tasks.

10. Surface electromyography signal obtained beneath an electrode is adequately representative of activity throughout the entire muscle.

11. A forward landing task is representative of a movement commonly employed in sport.

Delimitations

1. Only young healthy females and males with no history of lower extremity surgery or lower extremity injury within the immediately preceding six months were included in this study.

2. Femoral anteversion and hip ROM were measured using accepted clinical measurement methods.

3. Biomechanics were measured during the performance of single-leg and double-leg forward landings over a barrier normalized to 15% of each participant's height.

4. The mean of five trials is representative of a participant's single-leg forward landing strategy.

5. Surface electromyography electrode placed over the gluteus maximus is representative of hip external rotation and extension activation.

6. Surface electromyography electrode placed over the gluteus medius is representative of hip abduction activation.

7. Surface electromyography electrode placed over the adductor longus is representative of hip adduction activation.

7. For biomechanical testing, all participants wore standardized clothing and shoes to eliminate between-subject differences related to shoe-surface interactions.

Operational Definitions

Femoral anteversion: The angle (degrees) formed by the tibial diaphysis, as measured on a straight line between the tibial tubercle and the midpoint of the malleoli, and vertical when the participant is prone with the knee flexed to 90 degrees and the greater trochanter at its most lateral position as determined by palpation.

Hip internal rotation ROM (ROM_{IR}): The angle (degrees) formed by the tibial diaphysis, as measured on a straight line between the tibial tubercle and the midpoint of the malleoli, and vertical when the participant is prone with the knee flexed to 90 degrees and the femur is passively rotated internally until the point of initial sacral tilt as determined by palpation.

Hip external rotation ROM (ROM_{ER}): The angle (degrees) formed by the tibial diaphysis, as measured on a straight line between the tibial tubercle and the midpoint of the malleoli, and vertical when the participant is prone with the knee flexed to 90 degrees and the femur is passively rotated externally until the point of initial sacral tilt as determined by palpation.

Functional valgus collapse: A lower extremity movement pattern characterized by knee abduction, knee internal rotation, hip adduction, and hip internal rotation angles, and external joint moments associated with knee abduction and internal rotation, and hip adduction and internal rotation.

Single-leg forward landing: A functional task performed by jumping from two legs from a distance equal to 40% of the participant's height over a barrier equal to 15% of height and landing on the left leg.

Double-leg forward landing: A functional task performed by jumping from two legs from a distance equal to 40% of the participant's height over a barrier equal to 15% of height and landing on both legs.

Hip extension peak torque: The average maximum hip extension torque produced during two 5-second maximal isometric extension trials against a strap-assisted handheld dynamometer from a prone position with the hip in neutral and the knee flexed to 90°, normalized to the participant's moment arm (femur length).

Hip external rotation peak torque: The average maximum hip external rotation torque produced during two 5-second maximal isometric external rotation trials against a strap-assisted handheld dynamometer from a seated position with the hip and knee flexed to 90°, normalized to the participant's moment arm (tibial length).

Hip abduction peak torque: The average maximum hip abduction torque produced during two 5-second maximal isometric hip abduction trials against a strap-assisted handheld dynamometer from a side-lying position with the hip in 10-15° of extension and 10° of external rotation, normalized to the participant's moment arm (leg length).

Hip adduction peak torque: The average maximum hip adduction torque produced during two 5-second maximal isometric hip adduction trials against a strap-assisted handheld dynamometer from a supine position with the hip and knee extended, normalized to the participant's moment arm (leg length).

Gluteus maximus activation: Gluteus maximus muscle activation is being represented by surface electromyography signal obtained at a location one-third of the distance from the second sacral vertebrae and the greater trochanter. It is expressed as a percentage of EMG activation recorded during maximal voluntary isometric contraction (hip extension and hip external rotation peak torque, respectively).

Gluteus medius activation: Gluteus medius muscle activation is being represented by surface electromyography signal obtained at a location one-third the distance from the most lateral point of the iliac crest to the greater trochanter. It is expressed as a percentage of EMG activation recorded during maximal voluntary isometric contraction (hip abduction).

Adductor longus activation: Adductor longus muscle activation is being represented by surface electromyography signal obtained at a location one-third the distance from the left inferior angle of the pubic symphysis to the left medial femoral condyle. It is expressed as a percentage of EMG activation recorded during maximal voluntary isometric contraction (hip adduction).

Initial ground contact: The kinetic parameter defined by the moment at which the vertical ground reaction force (vGRF) exceeds 10N.

Landing phase: During a single-leg forward landing task, the phase commencing with initial ground contact and ending with peak knee flexion.

Healthy: An individual with 1) no history of lower extremity surgery, 2) no history of knee injury affecting ligamentous support or stability (including injuries to the ACL, MCL, PCL, LCL, medial meniscus, or lateral meniscus), 3) no history of lower extremity injury within the previous six months, 4) no presence of cardiovascular disease prohibiting moderate physical activity, and 5) no presence of vestibular condition affecting balance.

Independent Variables for Conventional Analyses

Femoral anteversion: Variable representing transverse structural alignment of the femur in relation to the pelvis.

Passive hip internal rotation range of motion (ROM_{IR}): Variable representing transverse capsular alignment of the femur as limited by the ischiofemoral ligament (Martin et al., 2008).

Passive hip external rotation range of motion (ROM_{ER}): Variable representing transverse capsular alignment of the femur as limited by the iliofemoral ligament (Martin et al., 2008).

Hip extension/external rotation peak torque: Variable representing maximum torque generation capability of the hip extensors and external rotators, respectively.

Hip abduction peak torque: Variable representing maximum torque generation capability of the hip abductors.

Gluteus maximus muscle activation : The peak RMS amplitude of the gluteus maximus from ground contact to maximal knee flexion during five trials of a forward landing normalized to the peak RMS EMG activation recorded during maximal voluntary isometric contraction (hip extension and hip external rotation, respectively).

Gluteus medius muscle activation : The peak RMS amplitude of the gluteus medius during five trials of a forward landing normalized to the peak RMS EMG activation recorded during maximal voluntary isometric contraction (hip abduction).

Sex: Female or male.

Landing task: Single-leg forward landing or double-leg forward landing.

Independent Variables for Statistical Parametric Mapping

Gluteus maximus muscle activation profile: A time series of RMS sEMG amplitude obtained from the gluteus maximus muscle over the course of the landing phase of five trials of a forward landing, interpolated and normalized to 101 data points.

Gluteus medius muscle activation profile: A time series of RMS sEMG amplitude obtained from the gluteus medius muscle over the course of the landing phase of five trials of a forward landing, interpolated and normalized to 101 data points.

Dependent Variables for Conventional Analyses

Initial Knee Abduction Angle: The frontal plane angle (°) formed by the tibia and femur at initial ground contact.

Peak Knee Abduction Angle: The maximum frontal plane angle (°) formed by the tibia and femur during the landing phase.

Knee Abduction Excursion: The difference, in degrees (°), between initial frontal plane knee angle and peak knee abduction angle.

Initial Knee Rotation Angle: The transverse plane angle (°) formed by the tibia and femur at initial ground contact.

Peak Knee Internal Rotation Angle: The maximum transverse plane angle (°) formed by the tibia and femur during the landing phase.

Knee Internal Rotation Excursion: The difference, in degrees (°), between initial transverse plane knee angle and peak knee internal rotation angle.

Initial Hip Adduction Angle: The frontal plane angle (°) formed by the femur relative to the pelvis at initial ground contact.

Peak Hip Adduction Angle: The maximum frontal plane angle (°) formed by the femur relative to the pelvis during the landing phase.

Hip Adduction Excursion: The difference, in degrees (°), between initial frontal plane hip angle and peak hip adduction angle.

Initial Hip Rotation Angle: The transverse plane angle (°) formed by the femur relative to the pelvis at initial ground contact.

Peak Hip Internal Rotation Angle: The maximum transverse plane angle (°) formed by the femur relative to the pelvis during the landing phase.

Hip Internal Rotation Excursion: The difference, in degrees (°), between initial transverse plane hip angle and peak hip internal rotation angle.

Peak Knee Abduction Moment: The maximum external joint moment acting about the anterior-posterior knee joint axis during the landing phase, normalized to height and weight ($N \cdot m \cdot BW^{-1} \cdot Ht^{-1}$).

Peak Knee Internal Rotation Moment: The maximum external joint moment acting about the axial knee joint axis during the landing phase, normalized to height and weight ($N \cdot m \cdot BW^{-1} \cdot Ht^{-1}$).

Peak Hip Adduction Moment: The maximum external joint moment acting about the anterior-posterior hip joint axis found during the landing phase, normalized to height and weight ($N \cdot m \cdot BW^{-1} \cdot Ht^{-1}$).

Peak Hip Internal Rotation Moment: The maximum external joint moment acting about the axial hip joint axis found during the landing phase, normalized to height and weight ($N \cdot m \cdot BW^{-1} \cdot Ht^{-1}$).

Dependent Variables for Statistical Parametric Mapping

Kinematic Knee Adduction/Abduction Profile: A time series of frontal plane knee adduction/abduction angles during the landing phase, interpolated and normalized to 101 data points.

Kinematic Knee Internal/External Rotation Profile: A time series of transverse plane knee internal/external rotation angles during the landing phase, interpolated and normalized to 101 data points.

Kinematic Hip Adduction/Abduction Profile: A time series of frontal plane hip adduction/abduction angles during the landing phase, interpolated and normalized to 101 data points.

Kinematic Hip Internal/External Rotation Profile: A time series of transverse plane hip internal/external rotation angles during the landing phase, interpolated and normalized to 101 data points.

Kinetic Knee Adduction/Abduction Profile: A time series of frontal plane knee external moments (normalized to height and weight) during the landing phase, interpolated and normalized to 101 data points.

Kinetic Knee Internal/External Rotation Profile: A time series of transverse plane knee external moments (normalized to height and weight) during the landing phase, interpolated and normalized to 101 data points.

Kinetic Hip Adduction/Abduction Profile: A time series of frontal plane hip external moments (normalized to height and weight) during the landing phase, interpolated and normalized to 101 data points.

Kinetic Hip Internal/External Rotation Profile: A time series of transverse plane hip external moments (normalized to height and weight) during the landing phase, interpolated and normalized to 101 data points.

CHAPTER II

REVIEW OF LITERATURE

Introduction

The purpose of this literature review is to give an overview of the evidence as it pertains to lumbo-pelvic-hip function and its relationship to functional valgus collapse and ACL injury risk. The aim is to provide a theoretical rationale for the proposed research questions. In keeping with this aim, evidence supporting a valgus collapse ACL injury mechanism will be discussed, as will evidence detailing structural, capsular, and neuromuscular components of the lumbo-pelvic-hip complex and their respective influences on functional valgus collapse and ACL injury risk. In so doing, I will highlight strengths of the literature base, as well as identify gaps to be addressed with future research. Additionally, methodological concerns within the current literature base will be discussed.

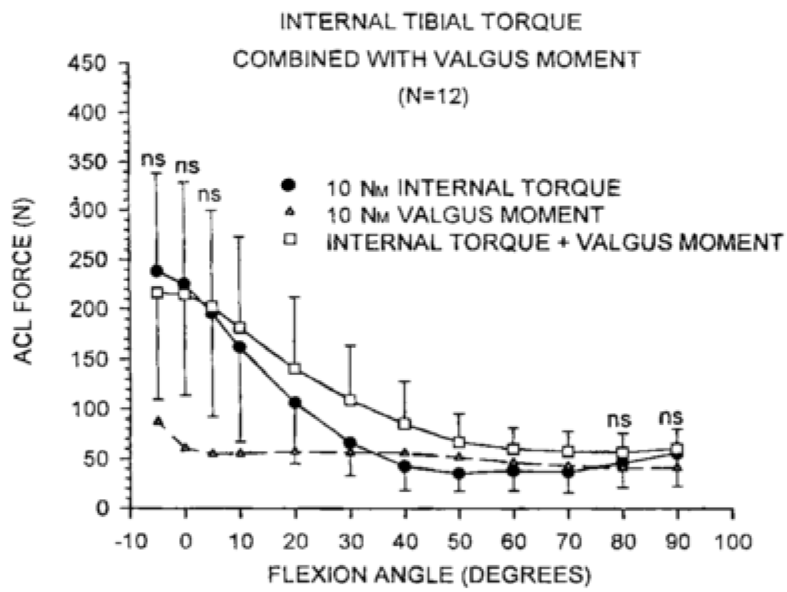
Contributors to ACL Strain

In order to investigate underlying causes of functional valgus collapse and their potential influences on ACL injury risk, an understanding of direct contributors to ACL strain is needed. Much research, using a variety of research designs, has been devoted to describing contributors to ACL strain. While cadaveric, in vitro studies provide much of the basis for current thought, retrospective videographic evidence and prospective studies describing potential predictors of ACL injury are also pertinent. Therefore, this section will provide a review of the literature base surrounding contributors to ACL strain and injury, and will be divided into three subsections: in vitro review, retrospective review, and prospective review.

In Vitro Review. The anterior cruciate ligament reaches anteriorly and medially from the medial border of the lateral femoral notch to the anteromedial tibial plateau. Because of this positioning, the ACL is thought to limit anterior tibial translation and internal tibial rotation. Indeed, much cadaveric work has been devoted to illustrate this concept. It has been shown that a pure tibial internal rotation torque can increase in-situ ACL strain by 117% and is capable of rupturing the ACL at a failure load of 37.4 kN, a load frequently produced during sport activity (Meyer, Baumer, Slade, Smith, & Haut, 2008; Y. Oh et al., 2011). Similarly, an anterior tibial force has been shown to increase ACL strain in vitro. Near full knee extension, force measured within the anteromedial bundle of the ligament reaches 180N, which equaled 150% of the applied anterior tibial force (Markolf et al., 1995). Furthermore, anterior tibial translation and internal tibial rotation are additive. The greatest amounts of strain within the ACL are induced by an anterior tibial force plus an internal tibial rotation force, with ACL forces in the anteromedial bundle reported to reach nearly 300N in magnitude (Markolf et al., 1995). The additive nature of these movements is important, as these loads likely do not occur in isolation. Due to anatomical constraints, it is thought that anterior tibial translation and internal tibial rotation are coupled motions. For instance, in the event of a more severe lateral posterior tibial slope, the lateral tibial plateau is encouraged to translate anteriorly more than the medial tibial plateau relative to the femur during functional weight-bearing movement, thus inducing internal tibial rotation (Beynnon et al., 2014; Marouane, Shirazi-Adl, & Hashemi, 2015; Meyer & Haut, 2008; Y. K. Oh et al., 2012) and further straining the ACL.

In the presence of anterior tibial translation and internal tibial rotation coupling, the addition of a frontal plane valgus force has repeatedly been shown to further increase ACL strain, particularly at knee flexion angles less than 30 degrees (Berns, Hull, & Patterson, 1992; Fukuda et al., 2003; Kiapour et al., 2015; Shin, Chaudhari, & Andriacchi, 2011). In the presence of anterior tibial translation and internal tibial rotation under weight-bearing conditions, the knee is more susceptible to a valgus collapse. The axial load introduced in weight-bearing can compress the lateral compartment and tension the MCL, which can then function as an axis of rotation around which the lateral compartment can rotate medially, thus potentially forcing the knee into a valgus position. As the knee moves into greater valgus collapse, the lateral compartment is compressed further (Meyer & Haut, 2008). These events may be exacerbated by the presence of greater lateral posterior tibial slope, which can cause the femur to "fall off" the back of the lateral tibia, inducing even further internal tibial rotation and placing maximal strain on the ACL (Berns et al., 1992). However, it is interesting to note that in the absence of internal tibial rotation and anterior tibial force, a pure valgus force only minimally increases ACL strain, if at all (Berns et al., 1992; Markolf et al., 1995; Oh et al., 2011). The argument for a multiplanar injury mechanism is made stronger by evidence showing that the ACL is less robust to torque applied in the transverse plane than in the frontal plane (Kiapour et al., 2015; Y. K. Oh et al., 2012). Figure 1 shows that when using similar amounts of torque, a rotary force induces greater strain within the ligament than a frontal plane moment, and that these moments are additive at knee flexion angles of 20-50° (Markolf et al., 1995). This would indicate that tibial rotation and anterior translation may be necessary components for an injurious valgus force to occur. Without these components occurring concomitantly, an isolated valgus force may be rendered impotent.

Figure 2.1. A Comparison between Internal Tibial Torque and Valgus Moment as Contributors to In Vitro ACL Strain (Markolf et al., 1995)



Retrospective Review. Retrospective videographic studies consistently indicate the presence of valgus knee collapse during ACL injury (Boden, Torg, Knowles, & Hewett, 2009; Krosshaug et al., 2007), as represented in Figure 2. Compared with sex-matched controls, injured males and females progressively moved into greater valgus collapse, with the injured cohort displaying frontal plane knee angles ten degrees greater than uninjured controls at the assumed moment of injury (66ms after initial contact) (Boden et al., 2009; Hewett, Torg, & Boden, 2009). In further videographic evidence, up to 53% of females display visible knee valgus at time of injury, compared with 17% of males (Krosshaug et al., 2007). This may indicate that females are more likely to injure their ACLs via valgus collapse mechanisms, whereas males may be more prone to alternative, more sagittal plane, injury mechanisms (Quatman & Hewett, 2009).

Figure 2.2. Representative Real-Time Observation of a Female ACL Injury (Olsen, Myklebust, Engebretsen, & Bahr, 2004)

Prospective Review. There is also prospective evidence supporting the relationship between functional valgus collapse and ACL injury. In 2005, Hewett et al. screened 205 female adolescent athletes (aged 15-16) during preseason using 3D motion capture of a drop vertical jump. Of the 205 screened, 9 went on to sustain ACL tears. Participants with ruptured ACLs were reported to display knee valgus angles 8° greater than uninjured counterparts. The primary variable of interest was peak knee abduction moment, which was shown to predict ACL injury status with 78% specificity and 73% sensitivity (Hewett et al., 2005). This study does have limitations. Nine ACL-injured athletes is a relatively small sample size, made smaller by the presence of an extreme outlier. Removing this outlier substantially weakens the relationship between peak knee abduction moment and ACL injury. Secondly, peak knee abduction moment refers to a pure frontal plane force, and this runs counter to cadaveric work indicating that a pure frontal plane valgus torque isn't likely to injure the ACL (Berns et al., 1992; Markolf et al., 1995; Oh et al., 2011).

In a replication of Hewett's original work, 710 elite female soccer and handball athletes, aged 21±4 years, were also screened using the drop vertical jump and tracked for 1-4 years (Krosshaug et al., 2016). Forty-two noncontact ACL injuries occurred that were suitable for analysis. Medial knee displacement was statistically different between injured and non-injured groups, though the mean difference was only half a centimeter (OR=1.40). Peak knee abduction moment and knee valgus at initial contact were not statistically different between groups. This could possibly be explained by the reported reliability of these characteristics. In a subset of the sample, test-retest reliability of motion capture was measured with 1-4 years between sessions, and the reliability of peak knee abduction moment is quite poor (ICC=.25). Because the average time between data collection and injury was 1.5±1.3 years, this study cannot conclusively claim that peak knee abduction moment is or is not associated with ACL injury.

Another recent publication detailed the relationship between 2D knee separation during a drop jump and general knee injury (OKane et al., 2016). While not aiming to explicitly predict ACL injury, this study has relevant implications. The sample consisted of females aged 11-14. Interestingly, the postmenarchal females in this cohort had a relative risk ratio of 3.62, indicating that females with the 10% most extreme valgus angles at maximum knee flexion were 3.62 times more likely to sustain a knee injury, whereas no such relationship existed in premenarchal females (OKane et al., 2016).

Collectively, these prospective studies indicate that functional valgus collapse may have a mild to moderate influence on ACL injury risk. Taking into account the varying methodology and populations between studies, drawing definitive conclusions is not possible. However, one explanation may be that the effect of functional valgus collapse on ACL injury risk could be a function of maturity and skill level, creating an inverted-U phenomenon. In premenarchal females and in elite, well-trained females, functional valgus collapse may only have a minimal impact on ACL injury risk. In lesser trained adolescent or college age females, functional valgus collapse may pose more of a threat. Further research using more homogenous methods and populations is needed to examine this potential effect.

One theme consistent throughout this literature base is that valgus collapse entails more than pure frontal plane movement and moments. It represents a number of factors colliding at the knee and forming what we call dynamic, or functional, knee valgus. Valgus collapse is more of a lower extremity profile, rather than a single joint motion as a risk factor. While it includes the knee motions of anterior tibial translation, internal tibial rotation, and valgus torque (knee abduction), it also encompasses hip adduction and internal rotation (Berns et al., 1992; T E Hewett et al., 2009; Ireland, 1999; T Krosshaug et al., 2007). To that end, many investigators have looked proximally to the hip for factors contributing to functional valgus collapse.

Hip-Knee Coupling

Globally, it is accepted that movement is generated proximally and transferred distally (Duval et al., 2010; Khamis & Yizhar, 2007; Reiman, Bolgla, & Lorenz, 2009). Accordingly, internal rotation and adduction of the femur is followed by internal tibial rotation and knee abduction. The reverse is also true: external femoral rotation predisposes one to external tibial rotation (Duval et al., 2010; Khamis & Yizhar, 2007). From this line of thinking has come the idea that the hip may be a key to better understanding knee motion. Paired with the dual-joint nature of functional valgus collapse, this has given birth to a body of literature detailing the ways in which hip movement couples with knee motion to potentially influence ACL injury risk.

Sagittal Plane. There has been evidence to suggest that decreased hip flexion upon landing may be a potentiator of knee valgus. Greater hip flexion upon landing and cutting allows for the absorption of ground reaction forces by contractile tissue. When sagittal plane hip flexion is insufficient to absorb energy, the resulting stiff-legged landing strategy is thought to cause the ground reaction force to spill over into the frontal plane, thus leading to ligamentous absorption of forces and potential functional valgus collapse (Hewett, Paterno, & Myer, 2002). Moreover, it has been suggested that insufficient hip flexion in the presence of ample knee flexion may increase the potential for shear forces between the femur and tibia (Hashemi et al., 2011), which in turn may increase anterior tibial translation, already described as a joint movement thought to contribute to knee valgus (S. J. Shultz & Schmitz, 2009b). These concepts have been corroborated in a number of research studies. Speaking to the presence of a landing strategy relying on ligamentous absorption of force, Schmitz et al (2007) reported that females who adopt a more stiff-legged landing than their male counterparts also elicit more ground reaction forces and absorb less energy with contractile tissue during a single-leg landing task, thus forcing inert (ligamentous) tissue to absorb the surplus energy, encouraging functional knee valgus (Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007). There is also evidence that makes a more direct association between decreased hip flexion and increased functional valgus collapse across various tasks. For instance, females previously shown to display greater dynamic valgus angles than male counterparts also exhibited decreased hip flexion during a side-step cutting maneuver (Pollard, Sigward, & Powers, 2007). No within-sex comparisons were made, thus making it difficult to determine how much of the differences were attributable to sex. However, the same group later made within-sex comparisons using a drop landing task and found that females displaying low hip flexion angles also exhibited increased knee valgus angles (Pollard, Sigward,

& Powers, 2010). Males were not included in this study, therefore it remains unknown if this mechanism is true in both sexes.

There are limitations within this body of evidence surrounding sagittal plane hip kinematics and their effect upon knee biomechanics. Specifically, the extent to which hip flexion couples with knee flexion hasn't been well described. Of the three studies reviewed, only one (Pollard, Sigward, & Powers, 2010) quantified both hip and knee kinematics concurrently. Examining hip and knee kinematics together would aid in determining the extent of hip-knee coupling in the sagittal plane, which may in turn provide an avenue for biomechanical intervention. Another limitation is the lack of within-sex research designs. Due to evidence suggesting knee valgus could be a sex-specific injury mechanism (Quatman & Hewett, 2009), it may not be appropriate to include both sexes in the same analyses. By comparing males to females, one cannot be certain whether knee movement strategies are truly due to proximal factors or of simply being male or female. As such, making between-sex or combined-sex comparisons may not be as beneficial as within-sex analyses.

Frontal Plane. Hip adduction is a chief component of functional valgus collapse (Ireland, 1999). It is accepted that increased hip adduction upon cutting and landing increases frontal plane knee load and in turn, valgus collapse (Timothy E Hewett & Myer, 2011; Imwalle, Myer, Ford, & Hewett, 2009). In a study of female soccer athletes, hip adduction was the only significant predictor of knee abduction during both 45 and 90 degree cuts (Imwalle et al., 2009), accounting for 25% of the variance in knee valgus angles during both cutting conditions. Also employing a 45 degree cutting task, Sigward & Powers (2007) found that females with excessive internal valgus moments demonstrated greater hip *abduction* at initial contact than females with lesser internal valgus moments (12.8±6.5 v. 7.7±6.4 degrees). Though this finding seems counterintuitive, it is likely due to the nature of the cutting task. In preparation for a side-step cut, the trunk moves toward the intended direction and away from the plant limb. As a result, the stance limb is abducted in preparation for push-off. Thus, during a sidestep cutting maneuver in which the trunk leans toward the planned direction, greater hip abduction at initial contact may be warranted in order to stay upright and successfully complete the maneuver. To confirm the task-specific nature of this strategy however, research using an alternative functional task is needed.

To more completely describe the extent of hip-knee coupling within the frontal plane, future research needs to include a greater variety of tasks. To date, the primary work in this area has been conducted within the purview of cutting maneuvers. To rule out the influence of trunk position, tasks such as drop jumps or forward landings would be beneficial to verify that hip-knee coupling relationships also exist in sagittal plane tasks and are comparable to those found in cutting tasks. Furthermore, this work is exclusively in females. Though it is necessary and useful to have within-female comparisons, thus avoiding sex-related confounds, it is noteworthy that these patterns have not been validated in an all-male cohort. As previously stated, exploring these relationships in a male cohort would aid in determining how much of this movement strategy is sex-specific.

Transverse Plane. Although empirical research is limited, hip internal rotation is also considered a key component of knee valgus collapse. During side-step and cutting maneuvers, females displayed increased hip internal rotation during the early deceleration phase of the task when compared to males (Imwalle et al., 2009; Pollard et al., 2007). Within sex, females possessing greater knee valgus also had more hip internal rotation at initial contact during a side-step cutting task than females displaying normal valgus alignment (Sigward & Powers, 2007), further suggesting that the hip and knee may be coupled joints.

Similar limitations exist in the transverse plane literature as in the previous sagittal and frontal plane literature. Females have been shown to have an increased propensity for dynamic hip internal rotation, which is a component of dynamic knee valgus. However, this pattern has been validated exclusively within cutting tasks. Perhaps it is a function of the task instead of faulty biomechanics specifically. Further work needs to explore the extent of hip-knee coupling during other tasks in which the ACL is typically injured, such as landing tasks.

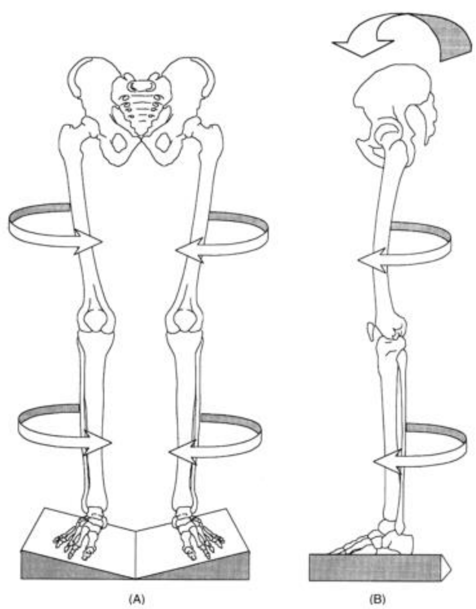
Summary. In summary, evidence for hip-knee coupling is mixed across planes. In the sagittal plane, research suggests that decreased dynamic hip flexion is associated with greater knee valgus, but this remains to be validated across tasks. Additionally, there is a need for sex-stratified research in this area. While evidence for hip-knee coupling is strongest in the frontal plane, cutting tasks have primarily been used to examine hip adduction's effects upon knee abduction. Though it's generally accepted that greater hip adduction is linked with more dynamic knee abduction, further research is needed to determine if its effect is task-dependent, or if it holds true in non-cutting tasks as well. In the transverse plane, while it is established that females exhibit more dynamic hip internal rotation than males, only one study observed that greater dynamic hip internal rotation is associated with an excessive valgus moment (Sigward & Powers, 2007b). Similar to the sagittal and frontal planes, more work is needed to confirm this transverse plane relationship in various tasks and in both sexes.

Factors Influencing Hip and Knee Function

From the perspective that the hip and knee move as a system coupled via the ground reaction force (Timothy E Hewett & Myer, 2011), factors that impact hip motion have the potential to impact knee motion, thus increase functional valgus collapse. To this end, there are multiple factors capable of influencing hip movement. They fall into three broad categories: hip structure and alignment, capsular restraints, and neuromuscular characteristics.

Bony Alignment. From a kinetic chain perspective, anterior pelvic tilt and femoral anteversion are two variants in the lumbo-pelvic-hip complex which have the potential to influence hip, and thus knee, motion. In theory, an increase in anterior pelvic tilt induces an acute internal femoral rotation, followed by internal tibial rotation (Figure 3) (Duval et al., 2010; Khamis & Yizhar, 2007). Acting along the same mechanistic lines, anteversion also represents an internally rotated femur and by extension an internally rotated tibia. There is evidence that both anterior pelvic tilt and an anteverted femur may contribute to functional valgus collapse.

Figure 2.3. Schematic Depicting Lower Extremity Kinetic Chain Mechanics as Theorized by Khamis & Yizhar (2007). Subtalar Pronation Sequentially Leads to Internal Tibial Rotation, Internal Femoral Rotation, and Finally Anterior Pelvic Tilt.



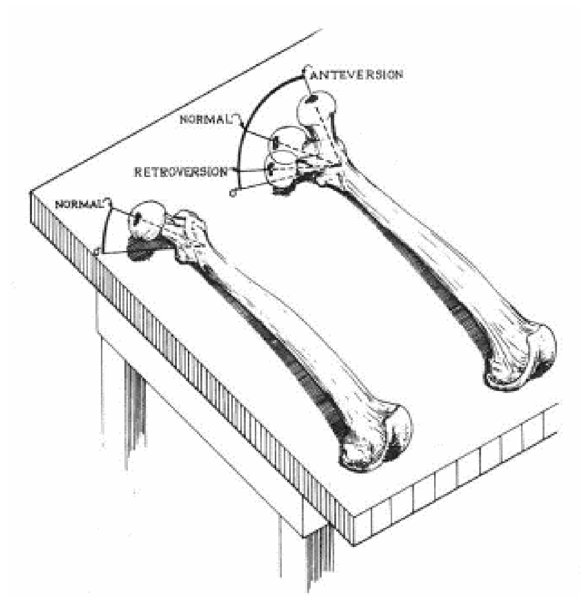
Anterior Pelvic Tilt. In theory, increased anterior pelvic tilt is thought to contribute to a valgus collapse mechanism. It is theorized that anterior pelvic tilt leads to limited femoral external rotation, or an increase in femoral internal rotation (Duval et al., 2010; Hruska, 1998), thereby increasing the potential for valgus collapse. In a retrospective chi-square analysis, an ACL-injured female cohort had significantly more anterior pelvic tilt than a healthy control group (Loudon et al., 1996), suggesting that Duval's theory, in which greater anterior pelvic tilt leads to internal femoral and tibial rotation, may assist in explaining a portion of ACL injury risk. Research conducted by Nguyen et al (2011) partially corroborated this theory, showing that less pelvic tilt (along with increased femoral anteversion, tibiofemoral angle, and navicular drop) predicted greater knee external rotation excursion during a single-leg squat. This supports Khamis & Yizhar's theory, in that both decreased anterior pelvic tilt and knee external rotation excursion are thought to be safer postures. However, knee rotation is referenced relative to the femur, so this finding is likely driven by the increase in femoral anteversion, which does not hold with Khamis & Yizhar's (2007) kinetic chain theory that an internal femoral rotation is followed by internal tibial rotation.

Potential confounds exist in these studies. Because pelvic tilt is largely dependent on posture and soft tissue restraints, it is possible that pelvic tilt was altered after ACL injury. Even though Loudon et al (1996) tested all participants within two years post ACL injury, it's impossible to verify that pelvic tilt didn't change during this interim due to alteration in muscle tensions. Also, only females were analyzed in Loudon et al's study, whereas Nguyen et al (2011) analyzed combined sexes, although sex was accounted for. Females are known to exhibit greater and more variable pelvic tilt than males (12±4.9° v. 8.7±4.1°) (Nguyen & Shultz, 2007). Because of the differences in variability, including both in these analyses may constitute a statistical confound by violating the assumption of homoscedasticity. Lastly, while ACL-injured females displayed greater anterior pelvic tilt in a chi-square analysis (p=.003) (Loudon, Jenkins, & Loudon, 1996), pelvic tilt was not predictive of group membership in a follow-up step-wise logistic regression within the same study, further questioning the veracity of the results.

In summary, evidence linking increased anterior pelvic tilt to hip and knee biomechanics and ACL injury is lacking. Future work should address the lack of cross-sectional, sex-stratified data as it relates to this topic. Because the pelvis is a foundational component in the kinetic chain, its influences may be multifactorial, encompassing other alignment characteristics, muscle stiffness and activation patterns, or postures. However, because pelvic tilt represents more of a postural characteristic than a structural one, its influences on lower extremity biomechanics may not be as stable as a true structural characteristic. Also, any influence it may have upon functional valgus collapse would manifest itself in a medially rotated femur. Therefore, accounting for a rotated femur should also account for pelvic tilt influences.

Femoral Anteversion. The articulation between the acetabulum and the head of the femur is variable in the transverse plane between individuals. Medial rotation of the femoral head within the acetabulum is called anteversion. This structurally rotated femur (Figure 4) causes a commensurate rotation at the knee. Given that dynamic hip internal rotation is a component of functional valgus collapse, it reasons that a static internally rotated femur may potentially bias the hip toward internal rotation and contribute to a greater dynamic knee valgus posture.

Figure 2.4. Comparison between Normal Femoral Neck Angle (left) and Femoral Anteversion and Retroversion (right) (Hoppenfield, 1976).



In spite of these theoretical connections, the evidence connecting hip anteversion to ACL injury is scant. Retrospectively, femoral anteversion did not discriminate between ACL-injured females and healthy females (Loudon et al., 1996). However, femoral anteversion was classified categorically as low (<8 degrees), normal (8-15 degrees), and high (>15 degrees), instead of as a continuous variable. Elsewhere, female means for femoral anteversion have been reported as being 14-18 degrees (Nguyen & Shultz, 2007, 2009; Shultz et al., 2009). Therefore, what was labeled as “normal” femoral anteversion could have actually been abnormal, thus leading to the preponderance of “normal” femoral anteversion measures (N=11 “normal,” N=0 “high,” N=9 “low”) (Loudon et al., 1996). The mis-categorization of femoral anteversion may have compromised the discriminatory power of the analysis, which may explain the nonsignificant results observed by Loudon et al (1996). Contrary to this study, another recent retrospective analysis in males (N=53) observed that for every 1 degree increase in femoral anteversion, ACL injury risk increased by 78% (Amraee et al., 2015). Thus, evidence linking femoral anteversion with ACL injury is mixed.

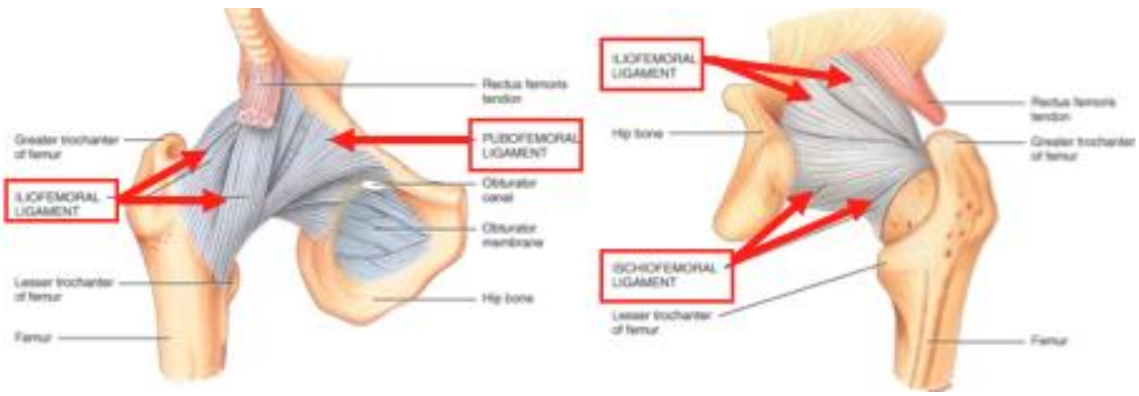
Cross-sectional research examining associations between femoral anteversion and components of functional valgus collapse is also scarce. In a multifactorial analysis using a single-leg squat and accounting for sex, greater femoral anteversion predicted greater hip internal rotation excursion and greater knee external rotation excursion (Nguyen et al., 2011), supporting the authors' hypothesis. A study published more recently by the same researchers confirmed the earlier results using a landing task. In this study, postures characterized by high femoral anteversion values maintained greater kinematic and kinetic valgus, as characterized by greater frontal plane knee kinematic valgus, greater external hip internal rotation moments, and greater external knee external rotation moments) throughout a drop-jump landing (A.-D. Nguyen et al., 2015). These indicate that femoral anteversion may exhibit an effect on hip and knee movement across various tasks.

Lastly, greater femoral anteversion has been shown to associate with increased anterior knee laxity (S. J. Shultz, Dudley, & Kong, 2012; S. J. Shultz et al., 2009), a characteristic prospectively shown to predict non-contact ACL injury in females (Uhorchak et al., 2003; Vacek et al., 2016). Increased anterior knee laxity increases the potential for anterior tibial translation during the transition from non-weight-bearing to weight-bearing (Daniel, Stone, Sachs, & Malcom, 1985; S. Shultz et al., 2006), and anterior tibial translation has already been established as a motion occurring concomitantly with functional valgus collapse. It is possible that an anteverted femur could place a chronic rotary strain on the ACL, which would increase anterior knee laxity, thus potentially predisposing one to greater anterior tibial translations and functional valgus collapse.

Summary. While a theoretical rationale exists for how the structural characteristics of anterior pelvic tilt and femoral anteversion influence functional valgus collapse and ACL injury risk, and while there is some evidence to support these mechanisms, there are still unknowns. Although there is substantial theoretical rationale regarding how anterior pelvic tilt influences hip and knee movement, compelling evidence is lacking. Kinetic chain theory suggests that pelvic tilt's potential influences on valgus collapse likely occur through a medially rotated femur. Thus, accounting for a torsioned femur should also account for pelvic tilt presence. Furthermore, the lone study analyzing anterior pelvic tilt's impact upon knee biomechanics uses a single-leg squat and a sex-combined cohort. Although sex was a covariate in this analysis, anterior pelvic tilt is more variable in females than in males (A.-D. Nguyen & Shultz, 2007). For this reason, completely separate analyses for males and females may be more appropriate when examining these variables. Similarly, the limitations in the evidence linking femoral anteversion to knee biomechanics are also a lack of sex-stratified research designs, as well as validation of this relationship across tasks.

Hip Capsular Constraints. The joint formed by the femoral head and the acetabulum, or the hip joint, is encased by a series of ligaments which form a cuff around the femoral head. The combination of these ligaments and the deep acetabulum create a stable joint, yet allow for substantial motion in all three planes of movement. Proximally, these ligaments insert along the lip of the acetabulum, just outside of the labrum. The femoral attachment site is along the intertrochanteric line anteriorly. Posteriorly, the cuff is more free, only partially covering the femoral head (Martin et al., 2008). More specifically, there are three primary ligaments of this cuff, each having its own role. These ligaments are depicted in Figure 5. The iliofemoral ligament is the largest of the three ligaments and is the primary component of the anterior capsule. It has two separate arms, a medial and a lateral arm (Martin et al., 2008). The medial arm functions to limit external rotation of the femur. The lateral arm has two functions. First, it limits adduction. Secondly, it limits internal rotation of the femur, particularly as the hip moves into extension. The ischiofemoral ligament forms the posterior capsule and serves as a limiter of internal rotation. Lastly, the pubofemoral ligament stretches along the inferior aspect of the hip. It functions to limit abduction (Martin et al., 2008).

Figure 2.5. Anatomical Arrangement of the Iliofemoral, Pubofemoral, and Ischiofemoral Ligaments. Left Side: Anterior View. Right Side: Posterior View.



It is also important to note that the envelope of passive hip ROM shifts as the hip moves in the sagittal plane. In hip flexion, more external rotation is possible. As the hip moves into extension, the envelope of passive ROM shifts toward internal rotation by approximately 20 degrees throughout the entire sagittal plane arc. Furthermore, the total arc of motion diminishes as the hip moves into extension (Martin et al., 2008; van Arkel, Amis, & Jeffers, 2015). Therefore, while the hip may be more loosely packed in flexion, a characteristic suggested to negatively affect joint control, an extended hip is biased toward internal rotation, which may increase one’s potential to exhibit functional valgus collapse (Pollard et al., 2007; Sigward & Powers, 2007b).

Sagittal Plane Passive Hip ROM. While the bulk of research surrounding passive hip range of motion has focused on the transverse plane, hip motion in the sagittal plane has received some attention. Given that the hamstrings and their posterior insertion on the tibia serve to protect the ACL, it is plausible that a more flexible hamstring could lend a differing degree of protection to the ACL. It has been argued that taut hamstrings may serve to prevent excessive anterior translation of the tibia (Cabaud & Rodkey, 1985). Furthermore, it has been suggested that the hamstrings may serve as anchors for the pelvis. In other words, more taut hamstrings may prevent the pelvis from moving into an anteriorly tilted position (Hruska, 1998), a posture previously described as potentially increasing functional valgus collapse and ACL injury risk.

Evidence supporting the potential impact of sagittal plane hip ROM on ACL injury is sparse. No significant differences in hamstring length were found between a female ACL-injured group and a matched control group (Loudon et al., 1996). Conversely, in a retrospective review of patient charts, ACL-injured females had significantly tighter hamstrings than their non-injured counterparts. Injured males in this study exhibited the opposite trend, displaying more flexible hamstrings when compared to healthy males (Harner, Paulos, Greenwald, Rosenberg, & Cooley, 1989). Reasons for these sex differences were not forthcoming. There were no cross-sectional studies identified which addressed the potential relationship between sagittal plane passive ROM and functional valgus collapse. Filling this gap may better elucidate the mechanism whereby sagittal plane ROM may influence ACL injury risk. However, because the hamstrings act primarily in the sagittal plane, their influences on functional valgus collapse would likely be indirect, acting via anterior pelvic tilt or increased anterior tibial translation. Therefore, measurement of sagittal plane mobility is not in the current proposal.

Transverse Plane Passive Hip ROM. Conventional thought states that an increase in passive hip internal rotation ROM (ROM_{IR}) leads to an increase in dynamic knee valgus by predisposing one toward greater dynamic hip internal rotation. Evidence supporting this theory is mixed. The lone cadaveric study analyzing the effect of ROM_{IR} upon ACL rupture revealed an inverse relationship between ROM_{IR} and peak strain within the anteromedial bundle of the ACL during a simulated pivot-shift (R^2=.91) (Beaulieu et al., 2014). This suggests that restricted ROM_{IR} may be more problematic than excessive ROM_{IR}. Of note, this study was specifically designed to mimic individuals with femoroacetabular impingement (FAI), a condition affecting the femoroacetabular articulation at its end range of motion. To mimic FAI, hard stops were set at the end ranges of motion. This is not representative of a typical athletic population, in which ROM is limited by soft tissue with a more pliable end-feel. Therefore, this study may not be generalizable to athletic populations in which ACL injuries are common.

Five retrospective studies addressing the relationship between passive hip ROM and ACL injury, all in males, were in agreement that a decrease in total ROM (driven by restricted ROM_{IR}) was associated with a history of ACL (Amraee, Alizadeh, Minoonejhad, Razi, & Amraee, 2015a; Bedi et al., 2014; João L Ellera Gomes, Palma, & Ruthner, 2014; Gomes, de Castro, & Becker, 2008; Tainaka et al., 2014). Of these studies, only one provided the length of time between sustaining the injury and ROM measurement (21-84 days post injury) (Amraee et al., 2015a), and only one of the five theorized how a restricted arc of motion may lead to ACL injury (Tainaka et al., 2014). This researcher suggested that insufficient transverse plane hip motion would necessitate that the knee provide the remainder of the motion needed to disperse rotary forces and safely complete the task. It was argued that in such a scenario, the tibia must internally rotate excessively, a motion known to strain the ACL (Berns et al., 1992; Fukuda et al., 2003; Markolf et al., 1995). A sixth retrospective study was identified which found no relationship between internal or external rotation hip ROM and ACL injury history (Hertel, Dorfman, & Braham, 2004). Despite the nearly homogenous conclusions drawn from the retrospective literature, substantial flaws in the research designs make it difficult to trust the robustness of these findings. For instance, one study lacked a control group (Amraee et al., 2015a), another lacked ROM variability, claiming that 95% of the study's 324 participants fell within one degree of the mean (Bedi et al., 2014), and other studies in this cohort fail to include measures of reliability (João L Ellera Gomes et al., 2014; Gomes et al., 2008). Also, further doubt is raised by the lack of a theoretical rationale for the majority of the conclusions within this body of evidence.

Contrary to the retrospective evidence, cross-sectional data supports the idea that excessive ROM_{IR} is deleterious, while greater amounts of ROM_{ER} may be beneficial. For instance, greater ROM_{ER} has been linked to decreased frontal plane knee excursion in females (r= -.40, p=.005) (Sigward et al., 2008). Moreover, females classified as having high ROM_{IR} remained in greater knee abduction throughout a landing task (Nguyen, Cone, Stevens, Schmitz, & Shultz, 2009). Combining these concepts that greater ROM_{ER} and lesser ROM_{IR} may be a desirable ROM profile, a variable termed AUHR (asymmetries of unilateral hip rotation; IR-ER) was developed. Higher AUHR values indicate more degrees of ROM_{IR} than ROM_{ER} within a given limb. This measure, along with peak hip abduction-external rotation torque and sex, explained 47% of the variance in knee abduction excursion during a single-leg landing task (Figure 2.6) (Howard, Fazio, Carl, et al., 2011).

Figure 2.6. Regression Equation Explaining 47% of Knee Abduction Excursion during a Single-Leg Landing (Howard, Fazio, Mattacola, et al., 2011).

Variable	Parameter Estimate, β	Standard Error	95% Confidence Interval	Standardized Estimate	*P* Value
Asymmetries of unilateral hip rotation	0.13	0.05	0.04, 0.22	0.336	.006
Peak hip abduction–external rotation torque	–5.41	2.17	–9.79, –1.03	–0.278	.02
Sex					
Male	Reference				
Female	5.19	1.39	2.39, 8.83	0.374	.001

[a] Knee abduction excursion = 11.63 + 0.13 (asymmetries of unilateral hip rotation) − 5.39 (peak hip abduction–external rotation torque) + 5.09 (sex = female). Adjusted $R^2 = 0.47$, $P < .001$.

It is important to note that each of the three aforementioned cross-sectional studies employed similar prone measurement methods, and each accounted for sex either in the research design or in the statistical approach, making comparisons possible. Though, given the increased variability common to females, including sex as a covariate may not be completely ideal. Upon comparison of the regression models by sex (Figure 6) (Howard, Fazio, Mattacola, et al., 2011), it seems possible that AUHR and hip abduction-external rotation torque may exert a greater influence in females than in males. However, further research would be needed to confirm this pattern.

An important distinction between the retrospective evidence, which argues that greater hip ROM (particularly ROM_{IR}) is desirable, and the cross-sectional evidence, which argues that less ROM_{IR} is desirable, is that the former advocates for greater *total* ROM, while the latter body of evidence emphasizes a decreased ROM_{IR} relative to ROM_{ER}. One possible way in which greater total ROM may influence ACL injury risk is through Generalized Joint Laxity (GJL). Moderate-to-strong correlations have been found between total hip ROM and GJL (r=.57) and between total hip ROM and hip laxity (r=.78) in a sex-combined cohort (Fan et al., 2014b). Therefore, total hip ROM may represent a laxity profile consistent with soft tissue restraints, and individuals with greater total hip ROM may also display laxity in other joints. Prospectively, individuals with GJL scores of 5 or higher, as determined by the Beighton and Horan Joint

Mobility Index (Beighton, Solomon, & Soskolne, 1973), are 2.8 times more likely to sustain an ACL injury (Uhorchak et al., 2003). Although this may be a plausible explanation linking total hip ROM to ACL injury, it would not agree with retrospective evidence advocating limited total hip ROM. Conversely, the reviewed cross-sectional evidence advocating decreased ROM_{IR}, either absolute or relative to ROM_{ER}, is more closely linked to knee biomechanics. Though only one study took into account the relative magnitude of internal rotation to external rotation (Howard, Fazio, Carl, et al., 2011), together the cross-sectional literature suggests that increased ROM_{IR} relative to ROM_{ER} may predispose an individual to greater valgus collapse by biasing the femur toward a more internally rotated posture.

Furthermore, in order to fully appreciate the potential mechanisms whereby hip ROM may exert influence over knee biomechanics, it is important to understand the potential relationship between relative ROM (ROM_{IR} / ROM_{ER}) and femoral anteversion. There is evidence to suggest a strong relationship between relative ROM and femoral anteversion (Howard, Fazio, Carl, et al., 2011; J Nyland, Kuzemchek, Parks, & Caborn, 2004), such that greater femoral anteversion may result in greater ROM_{IR}. During childhood, this is understood to be the case, as hip ROM_{IR} and femoral anteversion display correlations of 0.80 (Kozic et al., 1997). This relationship has not been quantified in adults, but is a possible factor influencing one's relative ROM. Throughout maturation, it is known that females retain greater anteversion than do males (S. J. Shultz, Nguyen, & Schmitz, 2008). It is also true that females possess greater ROM_{IR} (Fan, Copple, Tritsch, & Shultz, 2014a; Moreno-Pérez et al., 2015). Therefore, it is possible that disparate findings between sexes with regards to hip ROM are due to differences in femoral anteversion, which also may explain why injury risk associated with ROM_{IR} may not be the same in males and females.

In summary, a debate persists in the literature regarding the roles of ROM magnitude (total ROM) and ROM direction (relative ROM, or internal rotation bias), and their respective contributions to hip and knee biomechanics within each sex. Future work should consider both of these variables and seek to determine if each operates by directly influencing hip and knee biomechanics or through another mechanism, such as GJL. Additional gaps should also be addressed. Retrospective data consisted entirely of males, whereas cross-sectional evidence focused nearly exclusively in females, with Howard et al (2011) using a sex-combined cohort. The literature suggests that total ROM may play an important role in males (Gomes et al., 2014; Gomes et al., 2008). Conversely, ROM_{IR}, both absolute and relative, may play a more significant role in females (Nguyen et al., 2009; Sigward et al., 2008). Complementary analyses are needed in each sex to establish these potential sex-specific mechanisms. Additional work is needed exploring the role of hip ROM across different tasks. Each of the observational, cross-sectional studies used a sagittal plane landing task. Of the three, only Howard et al (2011) employed a single-leg landing. Whether these relationships hold during lateral tasks, such as a side-step cut, has yet to be investigated. Lastly, the relationship between relative ROM_{IR} and femoral anteversion needs to be established in adult male and female active populations.

Neuromuscular Function. ***Sagittal Plane.*** It has been suggested that sagittal plane hip strength and muscle activation patterns influence lower extremity biomechanics and ACL injury risk (D. R. Bell, Padua, & Clark, 2008; Khayambashi, Ghoddosi, Straub, & Powers, 2016a; Souza & Powers, 2009). In particular, the gluteus maximus is thought to play a particularly important role in the sagittal plane, as well as in the transverse plane, due to its dual action as a hip extensor and external rotator. This dual action is possible because the gluteus maximus originates along the sacrum and posterior iliac crest and inserts on the posterior aspect of the greater trochanter at the gluteal tuberosity. The fibers of the gluteus maximus are directed inferiorly and laterally from its origin. Because of this configuration, the gluteus maximus is thought to control downward deceleration during functional tasks, as well as limit hip internal rotation, which is important to lessen the potential for functional valgus collapse. However, evidence in support of the gluteus maximus' role during dynamic movement is inconclusive.

Sagittal plane gluteus maximus strength differences between sexes have been well documented, with females commonly exhibiting weaker hip extensors than males (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Homan et al., 2013a; Willson, Ireland, & Davis, 2006). For instance, normalized to height and weight, females exhibited decreased isometric hip extension strength when compared to males (13.21±0.8 v. 15.02±0.9 N*cm/kg) (Burnham et al., 2016). It has been demonstrated that females do not rely on the gluteus maximus to the same degree as males during activity, as was demonstrated by a more erect landing posture, decreased hip flexion angle, and a decreased eccentric hip extensor moment (Decker et al., 2003; T Krosshaug et al., 2007; Pollard et al., 2007; Schmitz et al., 2007). This upright posture profile is thought to be an ineffective method of allowing contractile tissue to absorb ground reaction forces, thus forcing ligamentous tissue to absorb these forces. Therefore, adopting an upright strategy may potentially lead to a functional valgus collapse mechanism (T. Hewett et al., 2002; A.-D. Nguyen et al., 2011; Zazulak, Straub, Medvecky, Avedisian, & Hewett, 2005).

While males and females differ in gluteus maximus strength, these findings alone are insufficient to suggest this as an ACL injury risk factor. Because there is evidence to suggest that females may injure their ACLs differently than males (Quatman & Hewett, 2009), research focusing on within-sex comparisons may be more appropriate. However, research analyzing the influence of hip extension strength on functional valgus collapse is weak. In both male and female cohorts, evidence revealed no associations between isometric hip extension strength and frontal plane knee excursion during a drop jump task and a forward lunge (Sigward et al., 2008; Thijs et al., 2007). The lack of positive findings suggests that isometric strength measures alone may not render a complete picture of the muscle's capability. Perhaps there is another neuromuscular component when, once accounted for, the relationship between hip extension and functional valgus collapse is strengthened. To address this idea, some researchers have explored the influences of muscle activation on functional valgus collapse, while co-varying for the muscle's strength (Homan et al., 2013a).

In addition to absolute torque producing capabilities of the hip extensors, how one utilizes, or activates, the gluteus maximus may also contribute to functional control of the hip. As such, measures of muscle activation may serve to highlight the difference between absolute torque generation capability and relative torque generation during a dynamic task. To illustrate, decreased gluteus maximus activation led to *decreased* knee valgus in a single-leg squat (Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011). While this would seem contrary to the hypothesis that greater muscle activation would better control hip motion, such findings are difficult to interpret based on activation alone. While hip extensor strength was not reported in this study, it is possible that individuals displaying decreased dynamic knee valgus also possessed stronger hip extensors, thus needing to recruit a *smaller* proportion of their available strength to effectively complete the task. This idea of less activation being indicative of a more efficient muscle will be a recurring concept in the discussion of all three planes of motion.

A limitation of this evidence is that it focuses primarily on absolute torque generation capability, largely excluding muscle activation considerations. Using EMG may help delineate between the roles of absolute torque capability and actual muscle activation during the task. Together, these two variables may determine the actual torque being produced during the task, and how it compares with the overall capability of the muscle. Additionally, future work should keep in mind the dual nature of the gluteus maximus. While it is a prime hip extensor, it is also a powerful external rotator. MVIC measurement of the gluteus maximus should consider this multiplanar configuration. Given the dual action of this muscle, normalizing EMG amplitude to extension strength versus external rotation strength could render differing results.

Frontal Plane. The gluteus medius is the primary muscle abducting the hip. It originates along the lateral surface of the ilium, just inferior to the iliac crest, and inserts on the lateral aspect of the greater trochanter. There is an assumption in the literature that an increased knee valgus angle could be the result of a weak or compromised gluteus medius, by permitting greater hip adduction (Homan, Norcross, Goerger, Prentice, & Blackburn, 2013b; Jacobs & Mattacola, 2005). Evidence indicates that insufficient hip abductor strength allows the hip to adduct during functional tasks, thereby increasing frontal plane knee load and subsequent knee valgus (Homan et al., 2013a; Jacobs et al., 2007; Mendiguchia, Ford, Quatman, Alentorn-Geli, & Hewett, 2011; Zazulak et al., 2005). Because frontal plane hip strength is thought to have a more direct relationship to functional valgus collapse than sagittal plane hip strength, more research has been devoted to the frontal plane.

Prospectively, there is moderate evidence to suggest a relationship between decreased frontal plane strength and increased risk of ACL injury. A recent prospective study showed that isometric strength of the hip abductors and external rotators separately predicted ACL injury (Khayambashi et al., 2016a). For every unit decrease in hip abduction strength (1 unit=1% body weight), athletes stood a 12% greater chance of sustaining an ACL injury. The only other available prospective study reported that decreased frontal and transverse plane hip strength were associated with all lower extremity injury (not specific to ACL injuries) in basketball and track & field athletes (Leetun, Ireland, Willson, Ballantyne, & Davis, 2004).

Cross-sectional evidence reporting significant relationships between frontal plane muscle strength and activation and lower extremity biomechanics is weak. No relationships were observed in sex-combined cohorts analyzing single-leg squats and double-legged landings (Homan et al., 2013a; A.-D. Nguyen et al., 2011). However, a single-leg landing task revealed a marginal association in females, as greater hip abduction torque moderately correlated with lesser knee valgus angles (r= -.35, N=15) (Jacobs et al., 2007). Using a single-leg step-down, isometric hip abduction strength was predictive of 2D frontal-plane knee valgus angle (r=.455) (Hollman et al., 2009a).

Similar to sagittal plane neuromuscular evidence, one potential explanation for these lackluster findings is that only one of the studies accounted for both strength and muscle activation (Hollman et al., 2009b). In the presence of weakened or compromised hip abductors, it is hypothesized that a heightened compensatory neural drive to the hip abductors is necessary to safely execute a task (Homan et al., 2013), ultimately resulting in a landing profile similar to those with stronger hip abductors. Thus, individuals with weaker hip abductors may be required to use a greater percentage of their MVIC during squatting and landing tasks (Homan et al., 2013; Nguyen et al., 2011), suggesting that decreased strength paired with greater, yet inadequate, muscle activation may set one up for greater functional valgus collapse. An individual displaying this profile would consistently operate near maximum capacity and may likely fatigue quickly, or may not be able to generate sufficient torque, even at high activation levels.

The biomechanical consequences of hip abductor weakness may become more apparent under fatiguing conditions. If an individual can complete a task using a smaller percentage of their hip abductor strength, it would reason that over the course of an athletic event, the gluteus medius would fatigue less, as there would be considerably more muscle fibers in reserve to recruit as needed. While it has been shown that a 30 second submaximal bout of hip abductor exercise increases peak hip adduction excursion during a single-leg forward landing in both sexes (r= -.31) (Jacobs et al., 2007), hip abductor muscle activation was not reported, so comparisons of muscle activation patterns cannot be made between the pre-fatigue and fatigued conditions.

Another potential explanation for the inconclusive findings focusing on the relationship between frontal plane muscle function and valgus collapse may be related to the tasks most commonly chosen in these studies. Researchers who used double-leg landings tasks or controlled, completely closed-chain single-leg tasks tended to observe no effects (Homan et al., 2013b; Anh-dung Nguyen et al., 2011). Studies that employed single-leg tasks, particularly those with a landing component, tended to observe significant effects (Hollman et al., 2009a; Jacobs et al., 2007). It is possible that a single-leg landing may tax the hip musculature to a greater extent, as it is more difficult to maintain a level pelvis during such movements.

Limitations include methodological inconsistencies across studies. These include: the use of isometric strength measures versus isotonic or isokinetic, the choice to analyze absolute torque measures alone, excluding muscle activation amplitude, and the types of populations studied. While an abundance of research uses isometric testing to obtain hip strength, this may not give a representative picture of a muscle's true capacity, in that it only yields information regarding muscle function at a specific joint angle. Isotonic or isokinetic testing, in which muscle function throughout a range of motion is obtained, may give more holistic information in an athletic population. However, isometric strength measures are more clinically feasible and easily obtained, as they do not require cumbersome and expensive equipment. For this reason, the majority of studies report isometric strength instead of isotonic or isokinetic. Another limitation is the choice to analyze strength data alone instead of also including muscle activation. In addition to absolute torque generating capacity, EMG amplitude relative to an MVIC could give a better understanding of how gluteus medius strength is used during a functional task (Homan et al., 2013b). Lastly, consensus between prospective research and cross-sectional studies is difficult due to population differences. While cross-sectional data primarily focused on adolescents (Homan et al., 2013b; Jacobs et al., 2007; A.-D. Nguyen et al., 2011), prospective studies included both mature, elite athletes and young, pre-pubertal females (Khayambashi, Ghoddosi, Straub, & Powers, 2016b). Given the inherent differences in experience, skill level, and training between these groups, comparisons may not be appropriate. Thus, it may be important to control for age and skill level when examining these relationships.

Based on available evidence, a relationship may exist between frontal plane hip strength and knee valgus during functional tasks. A small relationship was observed in single-leg landings (Jacobs et al., 2007). Consistently including a measure of muscle activation may help to clarify the relationship between frontal plane hip strength and activation and functional valgus collapse. Despite the inconclusiveness of cross-sectional research, prospective work is in agreement that both frontal and transverse plane hip strength are associated with injury (Khayambashi et al., 2016b; Leetun et al., 2004). The differences between cross-sectional work and prospective research could potentially result from population differences. Future work should seek to verify these mechanisms across all ages and skill levels.

Transverse Plane. Previously discussed as a hip extensor, the gluteus maximus also externally rotates the hip. This is due to its obliquely oriented muscle fibers, which are directed inferiorly and laterally from the later border of the sacrum. While thought to result in decreased dynamic hip flexion, a weak gluteus maximus is also suggested to permit greater hip internal rotation (Howard, Fazio, Carl, et al., 2011; Thijs et al., 2007; Willson et al., 2006). Because dynamic internal hip rotation is a component of knee valgus, the external rotators must work eccentrically to control this motion during the deceleration phase of a land or cut, thus working to prevent functional valgus collapse.

Prospectively, an injury odds ratio of 1.23 was reported for every percentage point decrease (as a % body weight) in hip external rotation strength (Khayambashi et al., 2016a). This means that for each percentage point of body weight decrease in absolute hip external rotation strength, the odds of sustaining an ACL injury increased by 23%. Also reporting external rotation strength as a percentage of body weight, a second prospective study showed significantly less isometric strength in basketball and track & field athletes who sustained lower extremity injuries than in healthy controls (Leetun et al., 2004).

Of the cross-sectional studies reviewed, three found no relationship between isometric hip external rotation strength and lower extremity biomechanics during double leg landing tasks and a 2D forward lunge analysis (Homan et al., 2013a; Sigward et al., 2008; Thijs et al., 2007). Thijs et al (2007), while not observing a significant relationship between isometric hip external rotation strength and valgus collapse, reported that a lesser ratio of external rotation strength to internal rotation strength was predictive of knee valgus (Thijs et al., 2007). For this reason, it may be beneficial to obtain strength and activation measures of antagonistic muscles such as the hip internal rotators and adductors. Another study observed a significant relationship between decreased peak isometric hip external rotation and increased 2D frontal plane knee angle ($r= .40$, $p=.004$) during a single leg squat (Willson et al., 2006). However, 2-dimensional motion capture is considered a poor surrogate for 3D analysis, as joint displacement correlations between 2D and 3D motion capture systems are reported to range from 0.12-0.34 (Olson, Chebny, Willson, Kernozek, & Straker, 2011). Lastly, it has twice been demonstrated, once in females and once in a sex-combined cohort, that decreased hip external rotation strength was associated with biomechanical variables often linked to functional valgus collapse (Howard, Fazio, Mattacola, et al., 2011; Lawrence, Kernozek, Miller, Torry, & Reuteman, 2008). Both of these studies analyzed 3D biomechanics during a single-leg drop landing task. Interestingly, Howard et al (2011) reported that peak isometric hip abduction-external rotation torque explained 9-16% of the variance in functional valgus collapse variables, while sex explained an additional 5-13% of the variance. This suggests that hip strength and sex may serve independent, and perhaps additive, roles in controlling knee motion. Furthermore, consistent with the frontal plane hip strength literature, studies that examined single-leg tasks revealed significant findings, while those using double-leg stance tasks tended to report null results. During single-leg stance, the stance limb's gluteal muscles are challenged to maintain a level pelvis while controlling the full body weight. These additional challenges may reveal deficiencies which might go unnoticed in a double-leg task.

There are limitations to consider. Although Lawrence et al (2008) reported greater external hip adduction moments in individuals with weaker external rotators, there was also a greater external knee *adduction* moment, contrary to the hypothesis. Additionally, no differences were revealed in frontal plane kinematics characteristic of knee valgus. Reasons for these inconsistent findings are unclear. Moreover, as discussed in the preceding section on frontal plane hip strength, there is a lack of studies examining the effects of isokinetic and isotonic strength on lower extremity biomechanics. This is likely because obtaining isometric strength is more clinically accessible and requires less equipment.

Summary. Based on this literature, the relationship between hip muscle function and functional valgus collapse is inconclusive. This relationship has not been established in the sagittal plane, and only a weak relationship has been documented in the frontal and transverse planes. While available evidence suggests that the use of single-leg landing tasks may better elucidate the effects of hip muscle function, further research comparing various tasks is needed to confirm this. The seeming lack of relationship may also be that both muscle strength and activation have not been consistently accounted for. It has been suggested that in weaker muscles, neural drive to the hip musculature must increase for safe completion of a task (Homan et al., 2013; Nguyen et al., 2011). Therefore, knowing the extent of muscle activation relative to the muscle's capability may better elucidate the role of hip muscle function.

Methodological Considerations

In light of the limitations noted within this body of research, it is appropriate to discuss pertinent methodological considerations and how these methodological choices may influence a study's outcome, thereby providing rationale for the current study's proposed methods. In brief, potential anatomical and functional interactions between components of the lumbo-pelvic-hip complex will be discussed, followed by task-related concerns, and finally statistical considerations.

Interactions Among Bony Alignment, Capsular Constraints, and Neuromuscular Characteristics. Components of the lumbo-pelvic-hip complex presented in this literature review are not isolated entities. Postural and structural alignment, capsular constraints, and neuromuscular function of hip and pelvic muscles all have the potential to influence hip control, and therefore likely interact with each other to influence lower extremity biomechanics. While these characteristics have been examined individually, the way in which these factors may interact with each other during functional movement has not yet been elucidated.

Though discussed previously in the section on transverse plane passive hip ROM, it is worth mentioning again the assumption held in the literature that femoral anteversion and hip

ROM, specifically ROM_{IR} relative to ROM_{ER}, are highly correlated (Howard, Fazio, Mattacola, et al., 2011; Kozic et al., 1997; John Nyland, Klein, & Caborn, 2010). While it has been demonstrated in children that femoral anteversion and ROM_{IR} are interchangeable (Kozic et al., 1997), due to the developmental nature of femoral anteversion, it is important to also establish this relationship in an adult population. This is supported by unpublished data from the Applied Neuromechanics Research Laboratory, in which correlations between femoral anteversion and ROM_{IR} and relative ROM_{IR} (ROM_{IR} – ROM_{ER}) were in .42 and .43 in females (N=112) and .64 and .65 in males (N=147), respectively. While these correlations are statistically significant, these data indicate that femoral anteversion may not be synonymous with ROM_{IR}, neither absolute ROM_{IR} nor relative to ROM_{ER}.

There is also evidence that transverse alignment of the femur, in the form of increased femoral anteversion and possibly increased ROM_{IR}, may influence gluteal function by lengthening the gluteal muscles' moment arm (Radin, 1979). Ordinarily, an elongated moment arm would result in a longer lever arm, thus a mechanical advantage. However, evidence has suggested that increased femoral anteversion and ROM_{IR} may separately be associated with decreased gluteal capability (Howard, Fazio, Mattacola, et al., 2011; Kaneko & Sakuraba, 2013a; J Nyland et al., 2004; Sigward et al., 2008). One possible explanation for this may be that muscle elongation resulting from increased femoral anteversion or ROM_{IR} may excessively stretch individual sarcomeres, eliminating the overlap between actin and myosin myofilaments. In such a case, the muscle could lose its ability to create cross-bridges. Examining these factors together may shed further light on the role of transverse plane femur alignment in functional valgus collapse. This argument is supported by the few studies analyzing gluteal function which did account for either femoral anteversion or hip ROM, consistently finding a relationship between gluteal function and knee biomechanics (Howard, Fazio, Mattacola, et al., 2011; Kaneko &

Sakuraba, 2013b; A.-D. Nguyen et al., 2011; Souza & Powers, 2009). However, no literature to date was identified which accounted for *both* femoral anteversion and hip ROM using a functional task to determine the role of gluteal function on lower extremity biomechanics. Doing so may help to further illuminate the influence of hip musculature on dynamic lower extremity motion and aid in the identification of underlying causes of valgus collapse, which can then be more directly targeted in injury prevention efforts.

Choice of Landing Task. Given that approximately 72% of ACL injuries occur during a single-leg stance (Barry P Boden et al., 2009), single-leg functional tasks may be more appropriate when seeking to identify ACL injury risk factors. Moreover, a task with an open chain "flight" phase may better help to provoke biomechanical effects stemming from the proximal segments. As already addressed, movement occurs in a proximal to distal direction. This is particularly true in an open kinetic chain system. In a closed kinetic chain system, the ground surface comes into play. As the foot makes contact with the ground, it moves into pronation, which in turn internally rotates the tibia (Duval et al., 2010; Khamis & Yizhar, 2007). Thus, force is propagated from distal to proximal in a fully closed kinetic chain system. However, because the majority of ACL injuries occur a mere 30-50 milliseconds after initial ground contact (T Krosshaug et al., 2007), it is possible that open chain kinetics are relevant to the initial loading of the joint. In other words, because ACL injuries occur so rapidly following initial ground contact, the hip and pelvis may still exert substantial influence over knee movement at the moment of injury. As the literature will attest, studies using single-leg tasks with open chain phases more often observe significant effects associated with lumbo-pelvic-hip movement and function (Hollman et al., 2009b; Homan et al., 2013b; Howard, Fazio, Mattacola, et al., 2011) than studies using double-leg tasks or single-leg closed kinetic chain tasks (A.-D. Nguyen et al., 2011; Sigward et al., 2008; Thijs et al., 2007; Willson et al., 2006). In the frontal and transverse planes in particular, single-leg open-chain tasks have proven more effective in eliciting gluteal muscle activation and identifying associated biomechanical effects (Hollman et al., 2009b; Homan et al., 2013b; Howard, Fazio, Mattacola, et al., 2011; Jacobs et al., 2007; Lawrence et al., 2008; A.-D. Nguyen et al., 2011; Sigward et al., 2008; Thijs et al., 2007; Willson et al., 2006). Despite this pattern, no studies have been identified comparing single-leg biomechanics with double-leg biomechanics, so this relationship cannot be directly confirmed.

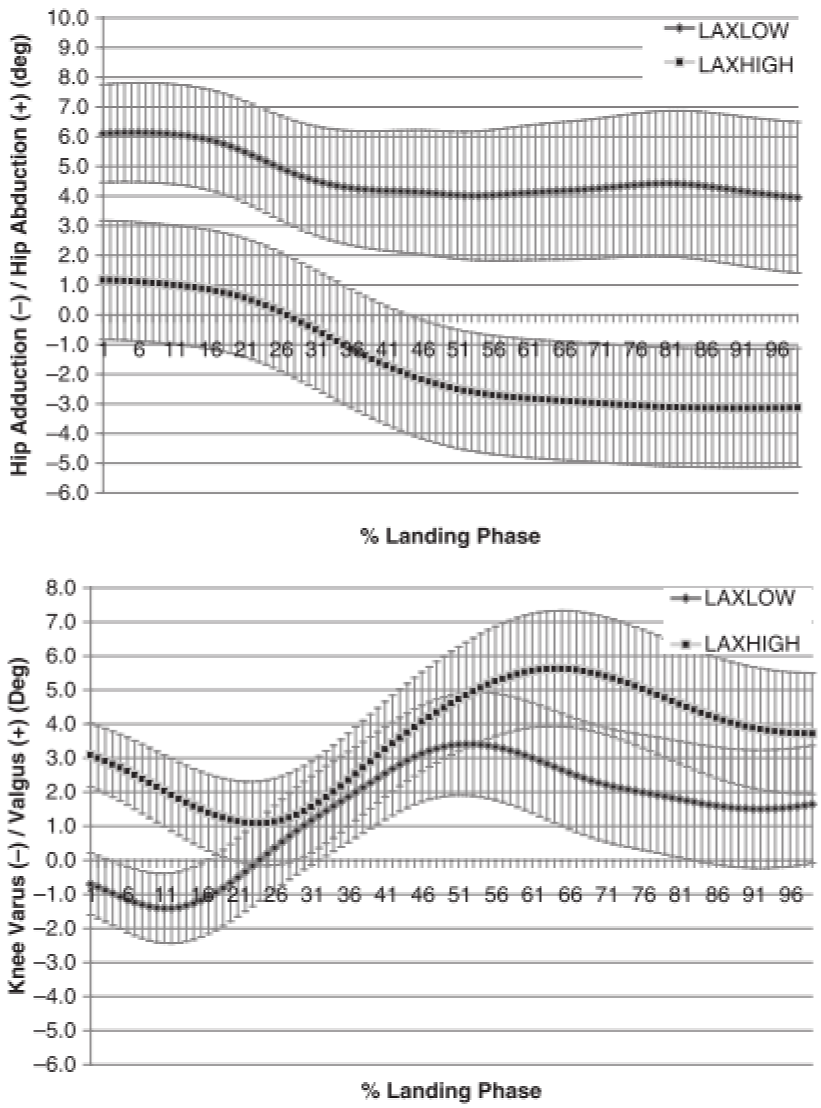
Two single-leg open-chain functional tasks previously used in the literature are the single-leg drop landing and the single-leg forward landing (Jacobs et al., 2007; Lawrence et al., 2008). In a single-leg drop landing, the subject drops off of two feet from a 40 cm box and lands on one leg (Lawrence et al., 2008). To perform a single-leg forward landing, the subject jumps with two legs over a barrier, landing on a single leg. The barrier and hop distance are normalized to 15% and 40% of the subject's height, respectively (Jacobs et al., 2007). While each of these tasks will theoretically place greater demands on the hip joint, the single-leg forward landing task is more akin to a game or practice situation, and to real-time injury mechanisms. In addition to maintaining a level pelvis, the horizontal propulsion needed to complete the single-leg forward landing task is an added challenge for the gluteals. Therefore, the functional task proposed for the current study is the single-leg forward landing.

Analysis Strategies. With the advent of motion-capture technology, time-series data are routinely obtained during functional tasks, such as a drop landing or side-step cut. However, conventional analysis of these biomechanical time-series data is often limited to analyzing discrete points on this continuum. Due to the need to adjust for type I error rate using conventional analyses, comparisons between many points along a time curve are ill-advised.

Thus, with a typical time series consisting of 3,000 data points (3 seconds x 1000 Hz sampling rate), it is common to discard nearly the entire series, retaining merely two or three singular data points that represent the joint motions and forces at discrete time points (e.g. at initial ground contact and maximum knee flexion). Though this analytic method is abundantly relied upon in the literature, it renders an incomplete picture of functional movement patterns. Specifically, it assumes linearity of joint loading, thus ignoring the possibility that prolonged joint loading, rate of loading, or erratic movement may impact one's injury risk.

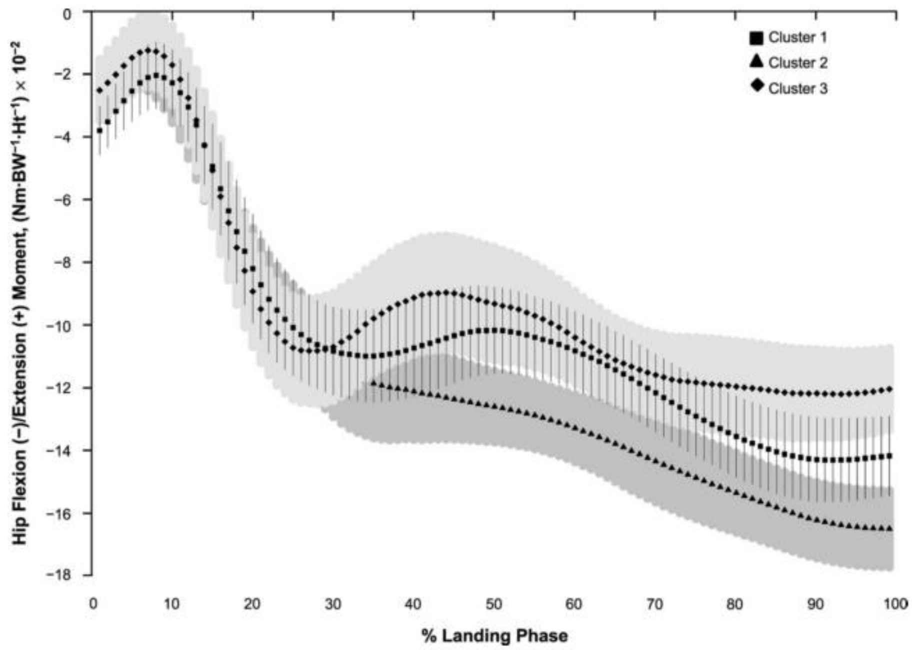
To address this problem, alternative statistical analyses more capable of handling entire time-series curves have been introduced in recent years. In particular, trend analysis with clustering, and statistical parametric mapping are beginning to exhibit more of a presence in the literature (Robinson, Vanrenterghem, & Pataky, 2015; S. J. Shultz & Schmitz, 2009a; Vanrenterghem, Venables, Pataky, & Robinson, 2012). To conduct a trend analysis, each time-series curve (e.g. knee abduction angle) must be normalized to a set number of data points, usually 101, stretching between two pre-determined events of interest, often initial ground contact and maximum knee flexion (S. J. Shultz & Schmitz, 2009a). The resultant time-series curves can then be ensemble averaged across subjects to yield a single group mean curve with standard deviation bars, examples of which are presented in Figure 7. Groups are determined by grouping subjects categorically, or clustering them, based on an independent variable(s) of interest. Group mean curves are then statistically compared using a modified repeated measures (trend) analysis, often conducted in SPSS. In this way, temporal comparisons cans be conducted, which renders a more complete picture of one's movement. For example, Shultz & Schmitz split subjects into high and low multiplanar laxity groups and analyzed these groups using a trend analysis (S. J. Shultz & Schmitz, 2009b). An excerpt from the results is seen below in Figure 7. As observed in the top graph, there were clear group differences throughout the entire landing phase. In the bottom graph, there were clear differences in the initial 15% of the landing phase, after which the groups behave similarly. Had only peak kinetics or kinematics been analyzed, as is custom, this pattern may have been missed.

Figure 2.7. Example of an Analysis Depicting Temporal Comparison between Participants with High and Low Knee Laxity (S. J. Shultz & Schmitz, 2009b).



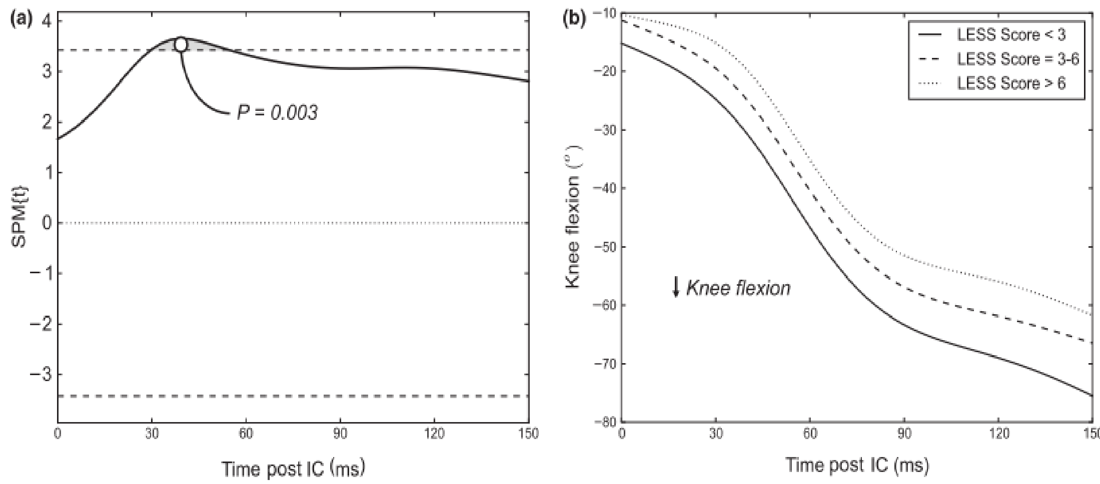
Taken a step further, it is also possible to group subjects on multiple variables using a cluster analysis before submitting the data to a trend analysis. For instance, subjects grouped according to similar lower extremity structural profiles (encompassing seven separate measures) constituted three distinct clusters, whose ensemble group means were then submitted to a trend analysis (A.-D. Nguyen et al., 2015). The first cluster was named the "internally rotated hip-valgus knee posture," and consisted of individuals with high pelvic tilt, femoral anteversion, quadriceps angle, tibiofemoral angle, and genu recurvatum, and low tibial torsion. Cluster 2, "neutral posture," was made up of individuals with average values for all seven measures. Lastly, Cluster 3 was named the "externally rotated knee-valgus knee posture" group. Participants in this cluster exhibited high pelvic tilt, quadriceps angle, tibiofemoral angle, and tibial torsion, average anteversion, and below average genu recurvatum. The 3-dimensional biomechanical time-series curves obtained during a drop-jump task were then ensemble averaged within each cluster and submitted to a trend analysis, similar to the study described in the preceding paragraph. A resulting graph from this study is shown in Figure 8. As can be seen in the graph, all three groups display similar patterns and values through approximately 30% of the landing phase, but then two groups diverge from the third, which continues linearly. Again, this pattern may have been overlooked if only initial and peak variables had been analyzed. Also, analyzing each of the seven clustering variables individually may have yielded nonsignificant results.

Figure 2.8. Example of an Analysis Depicting Temporal Comparison between Clusters of Subjects Displaying Different Profiles of Lower Extremity Structural Characteristics (A.-D. Nguyen et al., 2015).



In addition to trend analysis, statistical parametric mapping (SPM) has received recent attention in the literature (De Ridder, Willems, Vanrenterghem, Robinson, & Roosen, 2014; Dingenen et al., 2014; Pataky, Robinson, & Vanrenterghem, 2013). Originally developed from Random Field Theory to handle the enormous number of voxel to voxel comparisons necessary to analyze three dimensional brain MRIs (Pataky, 2012), only recently has SPM been adapted for use in biomechanics (Pataky et al., 2014, 2013). By circumventing the need to overly adjust for Type I error when making point-by-point comparisons, SPM allows for the analysis of entire time-series curves. This is accomplished by taking into account the inherent dependency of adjacent points on a time-series curve. The commonly used Bonferroni correction for familywise error rate halves the alpha level for each subsequent comparison, which is a severe correction that assumes each point to be independent of those adjacent. Meanwhile, SPM computes an alpha level using a method that accounts for the dependency of adjacent points on a time-series curve. An added feature of SPM is that correlative analyses are possible in addition to group comparisons, a feature not possible with a trend analysis. Since its introduction to the biomechanical community in 2012, both the SPM technique and its applications have been published in periodicals such as the Journal of Biomechanics, Computer Methods in Biomechanics, and Gait & Posture (Pataky, 2012; Pataky et al., 2014, 2013; Pataky, Vanrenterghem, & Robinson, 2015; Vanrenterghem et al., 2012). To illustrate its potential benefits, SPM was recently used to determine the relationship between LESS (Landing Error Scoring System) scores and knee flexion angles during an independent functional landing task (Fox et al., 2016). Higher LESS scores, indicative of more risky movement patterns, were significantly associated with reduced knee flexion. Interestingly, this relationship was only significant from 30-57 milliseconds after initial ground contact (Figure 9), roughly the same time frame during which ACL injuries are thought to occur (Fox et al., 2016; T Krosshaug et al., 2007). The authors further reported that conventional analyses using selected discrete variables yielded no statistically significant correlations. Therefore, this potentially critical piece of evidence would have been missed using only conventional statistical methods.

Figure 2.9. Descriptive and Inferential SPM Curves Describing the Relationship between Less Scores and Knee Flexion during a Functional Task (Fox, Bonacci, McLean, & Saunders, 2016).



Despite the promising results yielded by early use of methods such as trend analysis and SPM, there are substantial limitations with these techniques. The primary benefit of these methods is that a more complete picture of movement patterns can be gained by retaining the majority of a dataset. However, this can also be a drawback. By treating an entire time series curve as a single variable, it is overly cumbersome to control for a second variable, even with a powerful analysis such as SPM. For this reason, there is no acceptable method of including a covariate. By extension, this eliminates the explicit use of multiple regressions and ANCOVAs as potential statistical tools. However, should one wish to include multiple variables, it is possible to pre-emptively combine variables into a cluster or a factor score prior to use in an SPM analysis, similar examples of which were detailed previously in this section (Fox et al., 2016; A.-D. Nguyen et al., 2015). Lastly, explaining these techniques in an easily consumable manner is also a challenge and a current limitation to publishing such strategies. However, that barrier is slowly regressing as more and more researchers are becoming cognizant of the usefulness of including temporal, holistic analyses for biomechanical data. Eventually, this analysis may be more clinically meaningful than analyses relying solely on extracted initial and peak data points.

Conclusion

This literature review has provided an overview of the components of functional valgus collapse and their influence on ACL injury, the evidence supporting hip-knee coupling, as well as the ways in which the lumbo-pelvic-hip complex may influence gross lower extremity biomechanics in general, and valgus collapse in particular. There is evidence from in-vitro, retrospective, and prospective studies to indicate that functional valgus collapse (the combined motions of knee abduction, knee internal rotation, hip adduction, and hip internal rotation) influences ACL injury risk. Evidence from cadaveric studies indicates that internal tibial rotation and anterior tibial translation are additive and coupled motions (Markolf et al., 1995). In the presence of this coupling, a valgus force can further strain the ACL, though a valgus force alone only minimally increases ACL strain (Berns et al., 1992; Markolf et al., 1995; Y. Oh et al., 2011). As such, torque applied in the transverse plane seems to be a critical component for ACL rupture. Not only can a pure internal tibial rotational torque rupture the ligament, but the ACL is less robust to torque applied in the transverse plane than torque applied in the frontal plane (Kiapour et al., 2015; Y. K. Oh et al., 2012). These combined torques are thought to be present during functional valgus collapse, a movement pattern commonly observed on videographic footage of ACL injuries (Barry P Boden et al., 2009; Ireland, 1999; Tron Krosshaug et al., 2007). While retrospective evidence suggests that females may be at an increased risk of sustaining ACL injuries as a result of functional valgus collapse (Barry P Boden et al., 2009; Tron Krosshaug et al., 2007; Quatman & Hewett, 2009), prospective evidence in support of this injurious mechanism is mixed.

The hip and knee are thought to be coupled in their motion, and this coupling occurs via the GRF (Timothy E Hewett & Myer, 2011). Specifically, there is evidence to suggest that hip flexion, adduction, and internal rotation may be associated with knee flexion, abduction, and external rotation, respectively (Imwalle, Myer, Ford, & Hewett, 2009; Pollard, Sigward, Powers, et al., 2010; Pollard et al., 2007). Thus, controlling the motion and torques of the hip and pelvis may be a critical piece to controlling motion and torques at the knee. Yet, despite the amount of literature describing the relationship between hip and knee motion, there remain limitations and gaps. There is a need for more within-sex comparisons to determine if a valgus collapse mechanism is primarily a female concern. Also, much of the work has been conducted within the purview of side-step cutting tasks. More work is needed to establish hip-knee coupling across tasks, specifically sagittal plane, single-leg tasks. This is needed to confirm whether hip-knee coupling is a function of the task, or is inherent to the system.

There are a number of components of the lumbo-pelvic-hip complex that may potentially influence dynamic hip, and thus knee, function. These factors fall into three categories: postural and bony alignment, capsular constraints, and neuromuscular function. Because the kinetic chain theory indicates that anterior pelvic tilt leads to femoral internal rotation, which then leads to tibial internal rotation, the bony alignment characteristics of anterior pelvic tilt and femoral anteversion may impact hip and knee movement (Duval et al., 2010; Khamis & Yizhar, 2007). Greater amounts of anterior pelvic tilt and femoral anteversion have been empirically linked to greater functional valgus collapse (Amraee et al., 2015b; Loudon et al., 1996; Anh-dung Nguyen et al., 2011; Plastaras et al., 2015). Hip flexibility and capsular constraints also may influence lower extremity biomechanics. However, the mechanism for this influence remains unclear. While some studies indicate that greater ROM_{IR} is beneficial in avoiding functional valgus collapse and decreasing ACL injury risk, other studies hold that lesser ROM_{IR} is more advantageous for safer lower extremity biomechanics (Amraee et al., 2015b; Bedi et al., 2014; Joao L Ellera Gomes, Palma, & Ruthner, 2014; Gomes et al., 2008; Howard, Fazio, Mattacola, et al., 2011; A Nguyen et al., 2009; Sigward et al., 2008). Thirdly, neuromuscular function of the gluteus maximus and gluteus medius may have the potential to influence hip and knee control. While prospective evidence linking weak gluteals with ACL injury is compelling (Khayambashi et al., 2016b), cross-sectional data is mixed. It has been suggested that in addition to isometric hip muscle strength, muscle activation may be a pertinent variable. Specifically, decreased activation amplitude during a given functional task could indicate a stronger, more efficient muscle (Homan et al., 2013b; A.-D. Nguyen et al., 2015). Research operating under this assumption has yielded promising results thus far. Therefore, it may be important to include both absolute force generation capability and muscle activation amplitude in the research design. Finally, it is important to note that the aforementioned variables of bony alignment, capsular constraints, and neuromuscular function could have combined effects, but these have yet to be examined in combination. The gluteals insert along the greater trochanter of the femur. Therefore, varying amounts of femoral anteversion or hip ROM could displace the greater trochanter, which then may impact the gluteals' moment arm length, thus potentially influencing their neuromuscular capability and function (Radin, 1979). However, the way in which these factors combine to ultimately impact functional valgus collapse has not been elucidated.

Finally, methodological concerns present in previous literature should be addressed in future work. The first methodological concern is the chosen task. By eliminating the use of a drop box, and by recreating a single-leg landing, a single-leg forward landing may be a more realistic representation of real-life ACL injury mechanisms and may better elicit gluteal effects by placing greater demands on these stabilizing muscles. The second is the type of analytic approach used. Previous literature by and large has used conventional statistical analyses consisting of discrete time points, which may miss overarching movement patterns. The addition of tools such as trend analysis and Statistical Parametric Mapping may help to render more complete and holistic information regarding movement patterns during weight acceptance.

CHAPTER III

METHODS

Participants

A convenience sample of 45 female participants and 45 male participants was recruited, for a total of 90 participants. This sample size was based on an *a priori* power analyses and is adequate to detect moderate effect sizes, based on pilot data (see *Power Analysis* subsection on page 69) in a female cohort. As functional valgus collapse is thought to disproportionately affect females, this study is powered to examine female-specific mechanisms. However, because these mechanisms have not been elucidated in males, a corresponding male cohort was also recruited and examined for potential sex-specific mechanisms. To ensure a homogenous sample, specific inclusion criteria was 1) adults between the ages of 18 and 25 and 2) a score of two or more (at least "one time in a week") on categories 2-4 ("cutting, "decelerating", and "pivoting") of the Marx activity rating scale (see Appendix). Specific exclusion criteria was 1) any history of knee surgery, 2) any history of ligamentous or meniscal knee injury, 3) any history of lower extremity injury within the past 6 months, 4) history or diagnosis of a vestibular condition affecting balance, and 5) history or diagnosis of any cardiovascular condition precluding exercise. These exclusion criteria were in place because their presence has the potential to alter dynamic hip and knee biomechanics, or to incur unnecessary safety issues. Participants were largely recruited from the student population at the University of North Carolina Greensboro (UNCG), from where our lab routinely recruits participants. Each participant was compensated 20 dollars for their time.

Procedures

Participants reported to the Applied Neuromechanics Research Laboratory on UNCG's campus for a single testing session. After obtaining informed written consent, the following demographic information were obtained: height, weight, date of birth, and dominant stance limb. The PI provided a verbal explanation of the testing protocol to each participant, in addition to the explanation provided in the consent form. Each participant was asked to complete the following intake questionnaires: Physical Activity and Health History, Knee Outcome Survey (both the Activities of Daily Living Scale and the Sports Activities Scale), and the Marx Activity Rating Scale (Appendix A).

Anatomical Measures. Femoral anteversion and hip ROM were collected on the left leg. The left leg was chosen as representative based on evidence that femoral anteversion and hip ROM are comparable bilaterally within subjects (Hogg et al., in review; Shultz & Nguyen, 2007). Furthermore, using the same limb for all participants eliminated the need to relocate motion capture cameras for each participant. Also, the left leg was most often the self-selected stance limb. Thus, having the majority of participants land on their natural stance limb further emphasized the real-life nature of the task. Both femoral anteversion and hip ROM were measured prone with a standard inclinometer with the hip in neutral and the knee flexed to 90° (Magee, 1997). Considering cadaveric evidence which suggests that ligamentous configuration of the hip shifts as the hip moves into flexion, thus potentially altering ROM patterns, and considering that the hip typically remains in flexion during a landing phase, it may be more appropriate to measure hip ROM in 30° of hip flexion as opposed to 0°. However, analysis of unpublished data suggests that hip laxity variables obtained from neutral and 30° are highly correlated (internal rotation hip laxity in 0° and 30°: r=.952, external rotation hip laxity in 0° and

30°: r=.975, internal rotation : external rotation hip laxity in 0° and 30°: r= .955) (unpublished data from Fan et al., 2014, Applied Neuromechanics Research Laboratory, University of North Carolina Greensboro). As discussed previously, correlations between hip laxity and hip ROM are reported to be as high as .78 (Fan et al., 2014b). Therefore, because hip laxity measures obtained at 0° and 30° of hip flexion appear to be synonymous, hip ROM was only be obtained from 0° of hip flexion. To accomplish this, the examiner passively internally and externally rotated the lower leg while palpating the sacrum. At the point of initial sacral movement, the transverse angle formed by the tibial shaft and true vertical was measured as ROM_{IR} and ROM_{ER}, respectively. To measure femoral anteversion, the examiner rotated the lower leg while palpating the greater trochanter. When the greater trochanter was at its most lateral point, the transverse plane angle formed by the tibial shaft and true vertical was measured as femoral anteversion. Three trials were taken for femoral anteversion, internal and external rotation hip ROM and averaged for analysis. The PI had previously established good to excellent inter-day reliability with all of these measurements (Table 3.1).

Table 3.1. Intra-Rater Reliability Statistics for Hip Structural Measures.

Measure	$ICC_{2,3}$(SEM)
Femoral Anteversion	.92(1.2°)
Internal Rotation ROM	.97(1.6°)
External Rotation ROM	.85(3.3°)

Electromyography Sensor Placement. Surface electromyography (EMG) signals were acquired with double differential electrodes (Trigno Wireless Sensors, Delsys, Boston, MA) from the gluteus medius, gluteus maximus, and adductor longus during MVIC strength testing and performance of the single-leg forward landing. The adductor longus was included for the purpose of comparing its strength and activation to that of the gluteus medius, its antagonist. Prior to sensor placement, the skin was cleaned with an alcohol swab. The gluteus medius electrode was placed one-third the distance from the greater trochanter to the iliac crest (Rainoldi, Melchiorri, & Caruso, 2004). The electrode on the gluteus maximus was placed one-third the distance from the second sacral vertebrae to the greater trochanter (Rainoldi et al., 2004). The electrode on the adductor longus was placed one third the distance from the left inferior angle of the pubic symphysis to the medial femoral condyle (Lovell, Blanch, & Barnes, 2012). All electrodes were positioned parallel to muscle fiber orientation and secured with tape or prewrap. Proper positioning was verified with manual muscle testing and visual inspection of the EMG signal via the MotionMonitor oscilloscope.

Maximal Voluntary Isometric Contractions (MVICs). Because isometric strength measurement is more accessible in a clinical setting, and because there is no evidence advocating for the specific use of isotonic or isokinetic strength measures, isometric strength (MVICs) measures were obtained for the proposed study. Prior to obtaining MVICs, each participant completed a five minute warm up on a stationary bike at a self-selected pace. Following warm-up, MVICs of the gluteus medius, gluteus maximus, and adductor longus were obtained and used as maximum torque generation values, as well as for normalization of EMG amplitude. For all MVIC measures, a strap was used to secure the dynamometer in place and provide resistance for the participant. MVIC of the hip abductors (gluteus medius) was measured side-lying on the right side, with the left leg up, using a handheld dynamometer (Lafayette Instruments, Boston, MA). The left leg was placed in 10-15 degrees of hip extension and slightly externally rotated, thus isolating the gluteus medius. Maximal hip abduction was resisted by placing the lower edge of the dynamometer two inches proximal to the lateral knee joint line (Krause, Schlagel, Stember, Zoetewey, & Hollman, 2007). For MVIC measurement of the hip extensors (gluteus maximus), the participant was positioned prone with the knee bent to 90 degrees and maximally contracted into hip extension, with the dynamometer placed over the distal posterior thigh, two inches proximal to the joint line. Because the gluteus maximus has dual roles by also acting to externally rotate the hip, external rotation MVICs were also collected. To measure hip external rotation MVICs, the participant was seated at the end of a treatment table. The dynamometer was placed over the superior edge of the left medial malleolus. The femur was manually stabilized to minimize hip flexion and adduction during contraction, as most participants were inclined to compensate using these motions. Lastly, hip adduction MVICs were obtained from a supine position with the left leg extended and in zero degrees of hip abduction (Bohannon, 1986). The dynamometer was placed two inches proximal to the distal tip of the left medial malleolus. The right leg was stabilized and the participant was instructed to squeeze their legs together. Prior to collecting each MVIC, participants were familiarized to the measure and allowed a submaximal practice trial. All MVICs consisted of three 5-second trials, with 30 seconds rest between trials. To prevent an artificial spike in dynamometer output during collection, each participant was instructed to slowly increase their force, reaching maximum force production at three seconds of the five second trial. The PI had previously established reliability for these strap-assisted handheld dynamometry measures (Table 3.2).

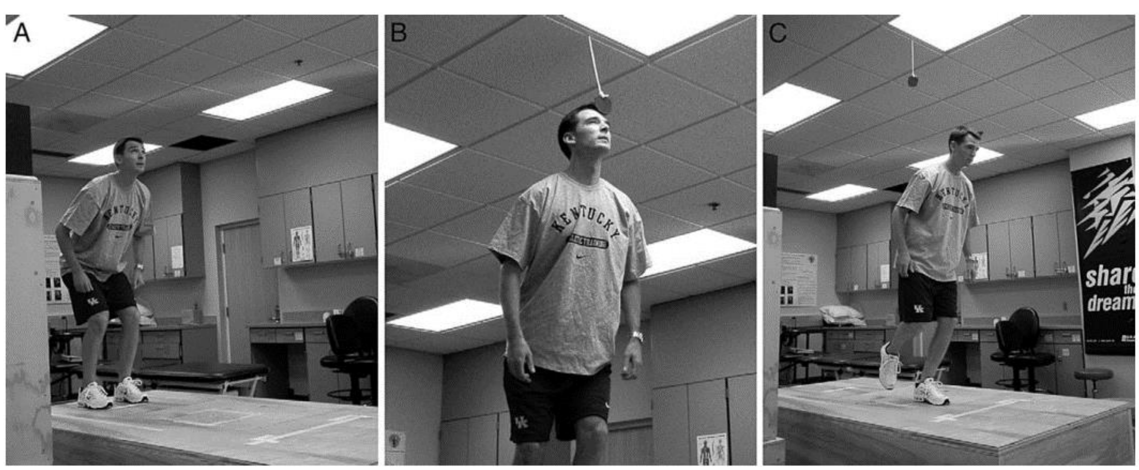
Table 3.2. Intra-Rater Reliability Statistics for Hip Strength Using a Handheld Dynamometer.

Measure	$ICC_{2,2}$(SEM)
Hip Abduction	.96(1.6)
Hip Extension	.76(3.4)

Single-Leg Forward Landing. Prior to digitization, participants were outfitted with standardized shoes (Adidas Uraha 2) to eliminate potential shoe-surface interactions. They were also asked to wear a shank sleeve made of thin material and outfitted with Velcro over their left calf. Participants were then be adorned with five marker clusters, each cluster with four optical LED markers, for a total of twenty markers. A marker cluster was placed at each of the following locations: lateral aspect of the left foot, lateral aspect of the left lower leg (mid-shaft), lateral left thigh (mid-shaft), the L5-S1 junction, and the postero-superior thorax (C7-T1 spinous processes). Clusters on the foot were held in place by adhesive backing and tape. Lower leg and thigh clusters were secured via Velcro to the standardized compression shorts and shank sleeve. The sacrum cluster was secured with double-sided adhesive tape and spray adhesive, and the thorax cluster was attached to a light harness, which also housed the battery pack for the LED sensors. Participants were then digitized using the MotionMonitor software (Innovative Sports Training, Chicago, IL). Joint centers for the knee and ankle were determined as the midway point between the medial and lateral femoral epicondyles and medial and lateral malleoli, respectively. The hip joint center was determined using the Bell method (A. L. Bell & Pedersen, 1989).

The functional task used to address the primary research questions was a single-leg forward landing onto an embedded forceplate (Type 4060-130; Bertec Corporation., Columbus, OH), measured with 3D motion capture (Figure 12). Participants were familiarized to the task prior to data collection and were allowed to practice the task until comfortable with the movement. Tape was placed on the ground at a distance equal to 40% of each participant's height away from the front edge of the forceplate. Participants were asked to stand behind the tape and jump from 2 legs over a foam barrier, landing on their left leg. The barrier's height was equal to 15% of the participant's height (Jacobs et al., 2007) and was placed halfway between the tape and the edge of the forceplate. The same instructions were given to each participant, and were as follows: “Please begin with your toes just behind the line. Take off of both feet, jump over the barrier, and land on your left leg, with your entire foot within the boundaries of the forceplate. Please make sure both feet clear the barrier without hooking around its edges. I will ask that you stick the landing for 2 seconds. Please keep your arms crossed against your chest throughout the task.” Trials were discarded if the participant double-hopped upon landing, hit the barrier, didn’t clear the barrier with both feet, didn’t land with the entire left foot on the forceplate, or used their contralateral limb for additional support. Five clean trials were collected and used for analysis. Additionally, because no studies were identified comparing single-leg landings to double-leg landings, five additional clean trials of a double-leg landing task were collected. The only difference between the single-leg and double-leg forward landing tasks was that the latter task entailed landing on both limbs, with the left leg fully on the forceplate and the right leg fully off. The order of the two landing tasks was counterbalanced.

Figure 3.1. Representation of a Single-Leg Forward Landing (Jacobs et al., 2007).

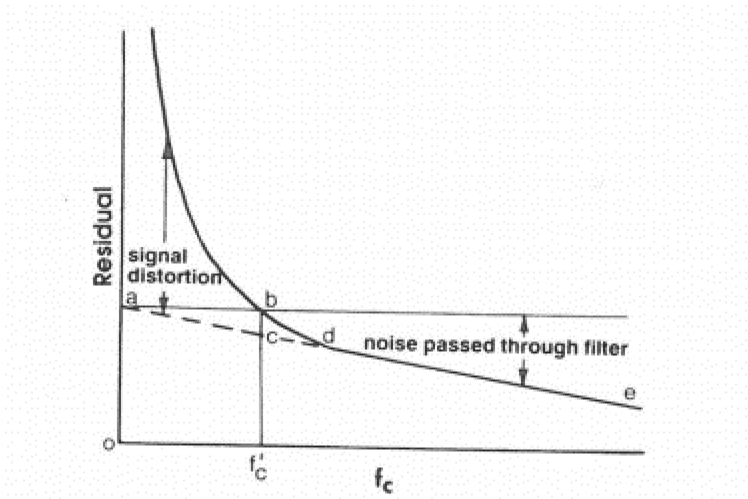


Data Sampling and Reduction

Maximal Voluntary Isometric Contractions. Peak torque data for each of the four MVIC conditions (hip extension, external rotation, abduction, and adduction) was recorded with a handheld dynamometer (Lafayette Instruments, Boston, MA) and collected for three trials of five seconds each. The highest two outputs for each condition were selected to represent the MVIC for each muscle, provided the two values are within ±10% CV. If the third trial registered the highest force output, then a fourth was collected. These criteria ensured that maximal effort was obtained, thus maintaining data integrity. A peak force, in Newtons, was recorded from the dynamometer for each trial, and the highest two, as identified by the previous stipulations, were averaged as the peak force. The peak force for each condition was then multiplied by the moment arm length (as determined by Dempster's data and accounting for placement of the dynamometer), then divided by participant body mass, resulting in a normalized torque, $N \cdot m \cdot kg^{-1}$. Surface electromyography (sEMG) data was collected concurrently with MVIC measurement. sEMG data was sampled at 1000 Hz and collected using EMGWorks (Delsys, Boston, MA) and exported to MATLAB for reduction using custom code. All sEMG data were filtered in MATLAB using a band-pass 20-350 Hz fourth-order, zero-lag, Butterworth filter with full-wave rectification. It was processed using a root mean squared (RMS) algorithm with a 25-millisecond time constant. The peak RMS sEMG amplitude was averaged across the same two trials selected for each MVIC condition, resulting in a peak EMG amplitude for each MVIC condition. These represented the maximum sEMG signal, and were used to normalize sEMG signal obtained during the single-leg and double-leg forward landings (% max EMG).

Single-Leg and Double-Leg Forward Landing Biomechanics. Biomechanical data were collected during each of the single-leg and double-leg forward landing trials. Motion capture began two seconds before initial ground contact, defined as the point at which the vertical ground reaction force exceeds 10N, and continued for three seconds after initial ground contact, for a total of five seconds. Kinematics were measured using an 8-camera optical LED system (Impulse, Phase Space; San Leandro, CA) at a sampling rate of 240 Hz. Kinetics were measured using a Bertec forceplate (Bertec Corporation, Columbus OH, USA). Kinetic and sEMG data were sampled at a rate of 1000 Hz. Kinematic and kinetic instrumentation were interfaced with MotionMonitor software and was manually synced by a pulse trigger during each trial. sEMG instrumentation was interfaced with EMGWorks (Delsys, Boston, MA) and was synced with kinematics and kinetics during processing using the built-in accelerometer in the sEMG sensor. To determine the appropriate filter for kinematic and kinetic data, a residual analysis was conducted on a subset of the total sample, and for each variable within this subset. To conduct the residual analysis, representative trials from four randomly selected participants were used. Each trial was exported under multiple conditions: raw, and at 2, 4, 6, 8, 10, 12, and 14 Hz. This yielded 8 "versions" of the same data series. From there, a sum of squares was computed for each non-raw version at each time point. This can be represented as $(\text{raw}_X\text{-filtered}_{yx})^2$, where *x* is the frame of data and *y* is the non-raw filtering frequency. Once the sums of squares are computed for each frequency, residuals can be obtained. For each filtering frequency, the residual is defined as the square root of the mean sum of squares across all time points (Winter, 1990). Finally, to determine the proper filtering frequency, the residuals were plotted, as shown below in Figure 11. The optimum low-pass filtering frequency is represented by $f_c{}'$. Both kinematic and kinetics for all analyses were filtered in MATLAB at the frequency determined by the residual analysis.

Figure 3.2. Graphic Representation of a Residual Analysis (Winter, 1990).



A segment-based coordinate system was used to define each body segment. The X-axis was defined as the anterior-posterior axis (adduction/abduction), the Y-axis was the distal-proximal axial axis (internal/external rotation), and the Z-axis was defined as the medial-lateral axis (flexion/extension). Motions for each joint were calculated using Euler's equations (Z Y' X") (Kadaba, Ranakrishnan, Wootten, Gainey, & Cochran, 1989). All flexions, adductions, and internal rotations were defined as positive angles. All extensions, abductions, and external rotations were defined as negative angles. Inverse dynamics was used to compute external moments for each joint (Gagnon & Gagnon, 1992) and were normalized to each participant's height (meters) and mass (kilograms). All data were then exported to MATLAB (MathWorks, Inc., Natwick, MA) for reduction using custom-written code.

For the conventional analyses, MATLAB code was written to extract all discrete variables from the exported data. Specifically, initial ground contact was defined as the point at which the vGRF exceeds 10 N. Initial ground contact marked the beginning of the landing phase. The end of the landing phase was the point of maximum knee flexion. Initial hip and knee angles (sagittal, frontal, and transverse) were defined as the respective joint angles at the moment of initial ground contact. The following peak joint angles were extracted: peak knee flexion, peak knee abduction, peak knee adduction, peak knee internal rotation, peak knee external rotation, peak hip flexion, peak hip abduction, peak hip adduction, peak hip internal rotation, and peak hip external rotation. These peaks were defined as the maximum respective joint angle occurring during the landing phase. Joint excursions were also calculated as the peak joint angle minus the initial joint angle, in degrees. Peak external moments were obtained for the hip and knee in each cardinal plane, and were defined as the maximum normalized moment during the landing phase. Lastly, peak RMS sEMG amplitude for the gluteus maximus, gluteus medius, and adductor longus obtained during the 150 ms prior to initial ground contact and during the landing phase were normalized to peak RMS sEMG amplitude obtained during MVIC testing and represented muscle pre-activation and activation during the landing phase.

For SPM analyses, MATLAB code was written to create time series curves of standardized length. The landing phase was extracted from all biomechanical data (kinematic, kinetic, and sEMG). Using a MATLAB interpolation function, time series curves of 101 data points, equally spaced, were generated for each variable listed in the preceding paragraph. Each curve extended from initial ground contact until the point of maximum knee flexion.

Statistical Approach

After processing and reduction in MATLAB, data were exported to Excel (Microsoft Corp., Redmond, WA). Data to be used for conventional analyses were organized and transferred to SPSS (IBM Corp, Armonk, NY), where statistical analyses were conducted. Data to be used for SPM analysis remained in Excel, where it was called in by MATLAB for SPM analysis.

Statistical significance for all analyses was 0.05 ($\alpha = 0.05$). Each hypothesis, along with its statistical approach, is detailed below.

Hypothesis 1a. Compared to males, females will exhibit greater functional valgus collapse, as exhibited by increased joint angles and external moments associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation. This pattern will be more pronounced in a single-leg forward landing than in a double-leg forward landing.

A 2x2 repeated-measures (sex by task) MANOVA was used to analyze differences by sex between single-leg and double-leg landing as it pertains to functional valgus collapse. Kinematic and kinetic variables associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation were examined. Kinematic variables were entered as initial contact angles, peak angles, and excursions (initial-peak). Kinetic variables were entered as peak external moments. MVICs for the gluteus maximus and gluteus medius were also examined, along with RMS muscle activation amplitude during pre-landing and during the landing phase.

Hypothesis 1b. Statistical Parametric Mapping T-tests, which examine biomechanical differences across the entire landing phase, will identify specific time points at which lower extremity biomechanics differ by task, and by sex, thus providing a more complete analysis than using discrete, singular time point variables.

This hypothesis was tested with pairwise SPM T-tests. Specifically, SPM T-tests comparing biomechanics between 1) single-leg and double-leg landing tasks and 2) between males and females were utilized. Dependent variables consisted of 101-point time series curves for each of the following: kinematic knee adduction/abduction profile (KA_{SPM}), kinematic knee internal/external rotation profile (KR_{SPM}), kinematic hip adduction/abduction profile (HA_{SpM}), kinematic hip internal/external rotation profile (HR_{SPM}), kinetic knee adduction/abduction profile (KAM_{SPM}), kinetic knee internal/external rotation profile (KRM_{SPM}), kinetic hip adduction/abduction profile (HAM_{SPM}), kinetic hip internal/external rotation profile (HRM_{SPM}), gluteus maximus muscle activation profile ($GMax_{SPM}$), and gluteus medius muscle activation profile ($GMed_{SPM}$). All SPM analyses were conducted using MATLAB code developed by Todd Pataky (Pataky, T., 2016, www.spm1d.org).

Hypothesis 2a. Increased femoral anteversion and ROM_{IR} and decreased ROM_{ER} will predict greater hip and knee movement toward functional valgus collapse during a single-leg forward landing task in females, as represented by greater peak knee abduction and internal rotation angles, peak hip adduction and internal rotation angles, peak external knee abduction moment, and peak external hip adduction moment.

To address this hypothesis, separate stepwise multiple linear regressions using the forward stepwise method were used. Specifically, the extent to which each component of functional valgus collapse is predicted by the independent variables of femoral anteversion, ROM_{IR}, and ROM_{ER} was examined. Dependent variables included frontal and transverse plane initial, peak, and excursion (initial-peak) values, as well as external peak moments for the hip and knee. Because both empirical evidence and functional anatomy indicate that the amount of potential knee valgus is dependent upon sagittal plane kinematics (Fukuda et al., 2003), peak hip and knee flexion were included as control variables. Prior to inspecting for statistical significance, each analysis was checked for assumption violations, specifically homogeneity of variance, multicollinearity, and leverage exhibited by an extreme outlier. To determine the relative contributions of each independent variable, partial correlations were inspected.

Hypothesis 2b. Collectively, increased femoral anteversion and ROM_{IR}, decreased ROM_{ER}, and increased gluteal activation with decreased hip strength will explain a greater portion of functional valgus collapse than femoral anteversion, hip ROM_{IR} and ROM_{ER} alone.

For this hypothesis, femoral anteversion, ROM_{IR}, and ROM_{ER} were entered into the first block of a regression analysis together. In the second block, strength and activation for each gluteal muscle were entered using the forward stepwise method. Dependent variables were the same as those listed for *Hypothesis 2a.* Due to reasons stated in *Hypothesis 2a*, and also because the gluteus maximus functions to eccentrically resist hip flexion during landing, hip and knee flexion were covariates in this analysis as well. Partial correlations were again inspected to determine the unique contribution of each variable to functional valgus collapse. Additionally, R squared changes were examined to determine the mediating effect of gluteal function on dynamic knee valgus. Kinematic and kinetic variables associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation were examined. Kinematic variables were entered as initial contact angles, peak angles, and excursions (initial-peak). Kinetic variables were entered as peak external moments.

Hypothesis 2c. A Statistical Parametric Mapping canonical correlation analysis, which takes into account the temporal nature of functional valgus collapse, will identify stronger relationships specific to points across the time series than will using conventional analyses with discrete, single time point variables.

To address this hypothesis, an SPM canonical correlation analysis was used to analyze the questions addressed by hypotheses 2a and 2b. MATLAB was used for all SPM analyses using code developed by Todd Pataky (Pataky, T., 2016, www.spm1d.org). To determine the influence of femoral anteversion and hip ROM upon temporal patterns of functional valgus collapse, three independent variables were used to create clusters which predicted time series curves of the following dependent variables: kinematic knee adduction/abduction profile (KA_{SPM}), kinematic knee internal/external rotation profile (KR_{SPM}), kinematic hip adduction/abduction profile (HA_{SpM}), kinematic hip internal/external rotation profile (HR_{SPM}), kinetic knee adduction/abduction profile (KAM_{SPM}), kinetic knee internal/external rotation profile (KRM_{SPM}), kinetic hip adduction/abduction profile (HAM_{SPM}), and kinetic hip internal/external rotation profile (HRM_{SPM}). To analyze the mediating effects of gluteal strength and activation using SPM, seven total independent variables were used: femoral anteversion, ROM_{IR}, ROM_{ER}, gluteus maximus MVIC, gluteus medius MVIC, gluteus maximus muscle activation profile ($GMax_{SPM}$), and gluteus medius muscle activation profile ($GMed_{SPM}$). These variables were condensed into clusters and then entered into an SPM canonical correlation analysis. The same dependent time curve variables were predicted. SPM output does not include R squared values, and there is no way to directly control for a variable. Therefore, unique contributions of each variable to the overall effect were roughly estimated via visual inspection of the data.

Power Analysis

An *a priori* power analysis was conducted using pilot data to determine the number of participants needed to achieve statistical significance. G*Power (Faul, et al., 2009) was used for all power analysis calculations. There is not currently a method for calculating an effect size using SPM. Therefore, power analyses are based on conventional statistical methods. From pilot data collected in healthy females aged 18-35, estimated effect sizes were calculated using peak knee abduction moment as the primary outcome variable, as literature has indicated this variable may be an important component of functional valgus collapse (Timothy E Hewett et al., 2005). Specifically, the following regression analyses were conducted to determine effect sizes (R^2):

1) To calculate an effect size for hypothesis 2a, a multiple linear regression (ENTER method) was used with three predictors and two control variables. After controlling for hip and knee flexion, the independent variables explained 21.8% of the variance in peak knee abduction moment.

2) To calculate an effect size for hypothesis 2b, a multiple linear regression (BACKWARD ELIMINATION method) was used. Of the seven predictor variables entered, the following two were retained: ROM_{ER} and gluteus maximus EMG amplitude, normalized to hip extension MVIC. These two variables explained 20.1% of the variance in peak knee abduction moment.

   a) Retaining three variables (adding gluteus medius EMG amplitude normalized to hip abduction MVIC) explains 26.7% of the variance in peak knee abduction moment.

   b) Retaining four variables (adding gluteus medius MVIC) explains 28.5% of the variance in peak knee abduction moment.

   c) Retaining five variables (adding ROM_{IR}) explains 30.0% of the variance in peak knee abduction moment.

3) Using the ENTER method, after controlling for ROM_{IR}, ROM_{ER}, and femoral anteversion, gluteus maximus and gluteus medius strength and activation accounted for an additional 8.2% of the variance in peak knee abduction moment.

Based on these effect sizes, an *a priori* power analysis was conducted for each hypothesis. For hypothesis 2a, including three predictors and two control variables (hip and knee flexion), a sample size of 44 was sufficient to detect a moderate effect size (R^2=.218) with power of 0.80. To appropriately power hypothesis 2b, a sample size of 42 was sufficient to detect an R^2 of .201 using two predictors and two control variables. Should three variables be retained in the final model (five total, including two control variables), a sample size of 35 was sufficient to detect an R^2 of .267. Should four variables be retained in the final model (six total, including two control variables), a sample size of 36 was sufficient to detect an R^2 of .285. Should five variables be retained in the final model (seven total, including two control variables), a sample size of 37 was sufficient to detect an R^2 of .300. While I did not anticipate that all nine predictors will remain in the final model (seven test predictors plus two control variables), Table 3 accounts for this possibility and details minimal detectable R^2 values using a given number of predictors and sample sizes. Therefore, 45 females were recruited for this study. To confirm this relationship in males, 45 males were also recruited, for a total of 90 participants.

Table 3.3. Minimal Detectable R^2 Values Given Number of Predictors and Sample Size. Note: Sample Size in Table Represents Single-Sex Cohort.

# of IVs	1	2	3	4	5	6	7	8	9
N=30	.22	.26	.30	.32	.35	.37	.39	.41	.44
N=35	.19	.23	.26	.29	.31	.32	.35	.36	.38
N=40	.17	.21	.23	.25	.28	.29	.31	.32	.34
N=45	.15	.19	.21	.23	.25	.26	.28	.29	.31

CHAPTER IV

MANUSCRIPT I. NEUROMECHANICAL SEX DIFFERENCES THROUGHOUT THE LANDING PHASES OF SINGLE AND DOUBLE-LEG FORWARD LANDING TASKS

Abstract

Context. Insufficiency of the hip muscles may be a contributing factor to dynamic knee valgus, but the role of hip musculature *throughout* functional movement has yet to be elucidated. As a first step toward a better understanding of the role of gluteal muscle activation in functional valgus collapse, there is a need to better characterize sex differences in neuromechanical control of the hip and knee during functionally relevant tasks.

Objective. To comprehensively examine sex-specific characteristics of functional valgus and neuromuscular control of the hip and knee throughout the landing phases of single-leg and double-leg landing tasks ($LAND_{SL}$ and $LAND_{DL}$) using two statistical techniques (General Linear Model and Statistical Parametric Mapping). We hypothesized that females would exhibit greater functional valgus collapse than males, despite utilizing greater gluteal activation (as a percentage of MVIC), and that these differences would be more pronounced during $LAND_{SL}$ than $LAND_{DL}$. Additionally, we anticipated that using Statistical Parametric Mapping (SPM) would allow us to better identify the time points when these sex and task differences were most pronounced.

Design. Cross-sectional.

Setting. Research laboratory

Patients or Other Participants. Forty-five females and forty-five males aged 18-25, with no history of lower extremity surgery or injury in the previous six months.

Intervention(s). Three-dimensional biomechanics and surface electromyography of the gluteus maximus and gluteus medius were obtained during performance of $LAND_{SL}$ and $LAND_{DL}$.

Main Outcome Measures. Frontal and transverse plane hip and knee initial joint angles, peak joint angles, joint excursions, external joint moments, and gluteus maximus and gluteus medius activation (% MVIC) were compared between sexes and tasks using 2x2 MANOVAs ($p<.05$). Time-series curves were generated for frontal and transverse plane hip and knee kinematics and kinetics, and for gluteus maximus and gluteus medius activation, and compared between sexes and tasks ($p<.05$) using separate SPM 2x2 ANOVAs.

Results. Sex differences in the frontal plane were task dependent, though females maintained greater absolute knee abduction and hip adduction throughout the landing phases. Sex by task interactions revealed that females landed with smaller knee adduction angles than males, particularly during $LAND_{SL}$ ([$LAND_{SL}$; F: 1.4±6.1°, M: 4.2±6.0°] [$LAND_{DL}$; F: 3.7±6.5°, M: 5.0±6.3°] $p=.03$), while females' knee abduction excursion was greater than males', particularly during $LAND_{DL}$ ([$LAND_{SL}$; F: 4.4±3.5°, M: 3.7±2.8°] [$LAND_{DL}$; F: 9.3±6.3°, M: 6.0±4.7°] $p=.01$). Across task, females displayed 4.1° greater peak knee abduction than males ($p=.002$), and SPM confirmed this was specific to 37-46% of the landing phase ($p=.05$). Females went through 1.0° more hip abduction than males (GLM $p=.05$), and used a smaller proportion of their gluteus maximus (35.8±21.8% MVIC v. 65.2±161.3% MVIC; $p=.01$) in both tasks.

Conclusions. Both analyses indicated that task demands are a significant moderator of sex differences in frontal plane hip and knee movement. Though females maintained greater knee abduction and hip adduction throughout the landing tasks, these differences were more profound during $LAND_{SL}$, potentially placing the knee in an even more precarious position with single leg tasks. Though males had higher gluteus medius activation to correspond with their greater hip abduction, further work is needed to examine the extent to which gluteus medius activation influences hip and knee control. Lastly, though SPM was not adept at identifying joint excursion effects, it was useful for defining the time parameters during which GLM main effects were valid. When seeking to examine biomechanical patterns such as functional valgus collapse, researchers should conscientiously choose the most appropriate task and analysis for their research questions.

Key Words. ACL, sex-specific, functional task, valgus, statistical parametric mapping

Introduction

Of the more than 350,000 anterior cruciate ligament (ACL) injuries that occur annually in the United States, an estimated 72% occur through non-contact mechanisms (Moses et al., 2012a; Wojtys & Brower, 2010). Functional valgus collapse, a movement pattern comprised of knee abduction, knee internal rotation, hip adduction, and hip internal rotation, is thought to increase the risk of sustaining a non-contact ACL injury (Hewett et al., 2005; Ireland, 1999). Retrospective videographic studies have consistently reported the presence of valgus collapse mechanisms during ACL injury (Boden, Torg, Knowles, & Hewett, 2009; Krosshaug et al., 2007). This mechanism is more commonly observed in females, where up to 53% of females are reported to display visible functional knee valgus at the time of injury, compared with 17% of males (Krosshaug et al., 2007). Additionally, females prospectively measured and found to have greater external knee abduction moments and smaller knee separation distances, both indicative of greater valgus collapse, were more likely to sustain ACL injuries than their healthy counterparts (Timothy E Hewett et al., 2005; OKane et al., 2016). These findings have led some to theorize that females may be disproportionately affected by valgus collapse mechanisms, whereas males may be more prone to alternative injury mechanisms (Quatman & Hewett, 2009). Yet despite these findings, factors contributing to the higher incidence of functional valgus collapse in females are not clear. Having this information is important as it will allow clinicians to more effectively tailor intervention strategies to sex and activity type.

As a first step toward that end, it is necessary to identify the best methods for assessing functional valgus collapse in women. One important consideration is the task under which functional knee valgus is evaluated. Given that approximately 72% of ACL injuries occur during a single-leg stance (Barry P Boden et al., 2009), it is reasonable to contend that single-leg tasks may present a greater challenge to maintaining safe lower extremity mechanics, thus better exposing factors contributing to this at-risk strategy. In fact, the use of more demanding, single-leg tasks have proven more effective in identifying sex differences than studies using less challenging double-leg tasks (Hollman et al., 2009b; Homan et al., 2013b; Howard, Fazio, Mattacola, et al., 2011; Jacobs et al., 2007; Lawrence et al., 2008; A.-D. Nguyen et al., 2011; Sigward et al., 2008; Thijs et al., 2007; Weinhandl, Irmischer, & Sievert, 2015; Willson et al., 2006). A single leg task places a greater demand on the ipsilateral hip abductors, which help to maintain pelvic stability and control knee motion, thus preventing functional valgus collapse (Howard, Fazio, Mattacola, et al., 2011; Jacobs et al., 2007). Moreover, females have often been shown to possess weaker hip abductors (Homan et al., 2013b; Howard, Fazio, Mattacola, et al., 2011; Jacobs et al., 2007), and require higher gluteus maximus and gluteus medius muscle activation percentages to perform single-leg activities than males (Hollman et al., 2009b; Zazulak et al., 2005). Hence, a single leg task may be preferred to not only magnify the potential for functional valgus collapse, but to better elucidate the role of the gluteal muscles in maintaining control of the hip and knee. Examining this methodological distinction and its impact on biomechanical sex differences is necessary so that future studies can unearth underlying factors that specifically contribute to, and ultimately effectively prevent, greater functional valgus collapse in females during single-leg activities where the ACL is more vulnerable.

Because functional valgus collapse is better described as a pattern of coupled joint motions, traditional statistical approaches that limit their analysis to isolated, discrete variables may not be ideal. While these approaches dominate the literature, extracting and analyzing only initial and peak position variables from biomechanical time-series data makes the assumption that movement is linear and that rate of loading is constant across participants. Not only is this assumption invalid, but it also fails to specifically elucidate the relevant biomechanical data near the 30-50 millisecond time window after intial contact during which ACL injuries are thought to occur (Carlson, Sheehan, & Boden, 2016; T Krosshaug et al., 2007). The few studies which have taken into account the temporal nature of functional valgus collapse were able to identify specific loading and timing differences between participants (Fox et al., 2016; A.-D. Nguyen et al., 2015; S. J. Shultz & Schmitz, 2009a), thus rendering a more complete picture of movement patterns. While promising, a limitation of these studies is that they do not provide a comprehensive description of how these patterns differ between sexes and in different tasks, nor how the information gained from a more integrated statistical approach compares to more conventional methods. Statistical Parametric Mapping is one such approach that allows for the comparison of complete time series curves by accounting for the dependency of adjacent time points in its calculation of the appropriate significance threshold, thus circumventing the conventional need to stringently adjust for Type I error rate. Thus, employing a more holistic statistical technique to our study of sex differences may provide more complete information of how males and females differ in their functional valgus collapse movement patterns.

Based on these stated gaps in the literature, the purpose of this study was to examine and compare sex-specific characteristics of functional valgus collapse (angles and external moments associated with greater knee abduction and internal rotation and greater hip adduction and internal rotation) and activation of the gluteal muscles throughout the landing phases of single-leg and double-leg landing tasks ($LAND_{SL}$ and $LAND_{DL}$) using two statistical approaches (GLM and SPM). Through traditional analyses of discrete variables (initial and peak angles and moments; gluteal activation), we hypothesized that compared to males, females would exhibit greater functional valgus collapse and would use a greater proportion of available torque production of their gluteal muscles, and that these sex differences would be more pronounced during the $LAND_{SL}$. Using Statistical Parametric Mapping, we expected to identify the specific time points during the landing phases where sex and task differences were most pronounced. Under the assumption that ACL injuries occur in the 30-50 milliseconds after initial ground contact, we specifically anticipated sex differences to be identified in this time window.

Methods

Participants. Forty-five female (20.1±1.7 yr, 165.2±7.6 cm, 68.6±13.1 kg) and forty-five male (20.7±2.0 yr, 177.7±8.5 cm, 82.8±16.3 kg) participants were recruited for a single testing session. To ensure a more homogenous sample, the study was limited to healthy young adults who were: 1) between the ages of 18 and 25 and 2) who scored two or more (at least "one time in a week") on categories 2-4 ("cutting," "decelerating," and "pivoting") of the Marx activity rating scale (see Appendix). Participants were excluded if they had: 1) any history of knee surgery, 2) any history of ligamentous or meniscal knee injury, 3) any history of lower extremity injury within the previous 6 months, 4) history or diagnosis of a vestibular condition affecting balance, and 5) history or diagnosis of any cardiovascular condition precluding exercise. Each participant provided written informed consent as approved by the university's Institutional Review Board. After obtaining written informed consent, each participant's height and weight were measured, followed by completion of the following intake questionnaires: Physical Activity and Health History Questionnaire, Knee Outcome Survey (both the Activities of Daily Living Scale and the Sports Activities Scale), and the Marx Activity Rating Scale (Appendix).

Surface Electromyography Instrumentation. Each participant was outfitted with surface electromyography (EMG) double differential electrodes (Trigno Wireless Sensors, Delsys, Boston, MA) to acquire signals from the gluteus medius and gluteus maximus during maximal strength testing and performance of the $LAND_{SL}$ and $LAND_{DL}$. Prior to sensor placement, the skin was cleaned with an alcohol swab. The gluteus medius electrode was placed one-third the distance from the iliac crest to the greater trochanter (Rainoldi et al., 2004). The electrode on the gluteus maximus was placed halfway between the second sacral vertebrae to the greater trochanter (Rainoldi et al., 2004). All electrodes were positioned parallel to muscle fiber orientation and secured with tape or prewrap. Proper positioning was verified with manual muscle testing (Starkey & Ryan, 2003) and visual inspection of the EMG signal using EMGWorks (Delsys, Boston, MA). EMG data were sampled at 1000Hz and were manually synced with kinematic and kinetic data during post-collection processing.

Maximal Voluntary Isometric Contractions (MVICs). Prior to obtaining MVICs, each participant completed a five minute warm up on a stationary bike at a self-selected pace. Following warm-up, MVICs of the gluteus maximus (hip extension) and gluteus medius (hip abduction) were obtained and used to document maximum torque production, as well as for normalization of EMG amplitude. For all MVIC measures, a strap was used to secure the dynamometer in place and to provide resistance for the participant. For MVIC measurement of the hip extensors, the participant was positioned prone with the knee bent to 90 degrees. With a handheld dynamometer (Lafayette Instruments, Boston, MA) placed over the posterior distal thigh two inches proximal to the joint line, the participant was asked to maximally contract into hip extension (Bohannon, 1986; Starkey & Ryan, 2003). Hip abduction MVICs were measured side-lying on the right side, with the left leg up. The left leg was placed in 10-15 degrees of hip extension and slightly externally rotated, thus isolating the gluteus medius. Maximal hip abduction was resisted by placing the lower edge of the dynamometer two inches proximal to the lateral knee joint line (Krause et al., 2007). Prior to collecting each MVIC, participants were familiarized to the measure and allowed up to three submaximal practice trials. Each condition consisted of three 5-second trials, with 30 seconds rest between trials. To prevent an artificial spike in dynamometer output during collection, each participant was instructed to slowly increase force output, reaching maximum force production during the third second of the five second trial. The PI had previously established reliability for strap-assisted handheld dynamometry [$ICC_{2,2}$(SEM); hip extension: .76(3.4kg); hip abduction: .96(1.6kg)].

Biomechanical Instrumentation. Prior to digitization for motion capture, participants were outfitted with standardized shoes (Adidas Uraha 2, Adidas AG, Herzogenaurach, Bavaria) to eliminate potential shoe-surface interactions. Participants were adorned with six marker clusters, placed at each of the following locations: lateral aspect of the left foot, lateral aspect of the left lower leg (mid-shaft), medial and lateral proximal tibial flares, the lateral left thigh (mid-shaft), the L5-S1 junction, and the postero-superior thorax (C7-T1 spinous processes). Participants were then digitized using the MotionMonitor software (Innovative Sports Training, Chicago, IL). Joint centers for the knee and ankle were defined as the midway point between the medial and lateral femoral condyles and medial and lateral malleoli, respectively. The hip joint center was determined using the Bell method (A. L. Bell & Pedersen, 1989). A segment-based coordinate system was used to define each body segment. The X-axis was defined as the anterior-posterior axis (adduction/abduction), the Y-axis was the distal-proximal axial axis (internal/external rotation), and the Z-axis was defined as the medial-lateral axis (flexion/extension). Motions for each joint were calculated using Euler's equations (Z Y' X") (Kadaba et al., 1989). Kinematic data were obtained with an 8-camera optical LED system (Impulse, Phase Space; San Leandro, CA) at a sampling rate of 240 Hz. Kinetic data were obtained using an embedded Bertec forceplate (Type 4060-130; Bertec Corporation, Columbus OH, USA) and were sampled at a rate of 1000 Hz. Kinematic and kinetic instrumentation were interfaced with MotionMonitor software and were manually synced by a pulse trigger during each trial.

Procedure for Single and Double-Leg Forward Landing. Participants were familiarized to each task prior to data collection and were allowed up to three practice trials. For both tasks, tape was placed on the ground at a distance equal to 40% of each participant's height away from the front edge of the forceplate. A foam barrier equal to 15% of the participant's height (Jacobs et al., 2007) was placed halfway between the tape and the edge of the forceplate. Instructions to participants were standardized. For the $LAND_{SL}$, participants were instructed to stand behind the tape and jump from 2 legs over the foam barrier, landing on their left foot. The procedure for the $LAND_{DL}$ was the same in every way except the participants landed on both feet, with the left foot landing entirely within the forceplate, and the right foot landing completely outside the forceplate. Trials were discarded if the participant double-hopped upon landing, hit the barrier, didn't clear the barrier with both feet, didn't land with the entire left foot on the forceplate, or used their contralateral limb for additional support. Five clean trials for each task were collected and used for analysis. The order of the two landing tasks was counterbalanced.

Data Handling and Processing. Motion capture began two seconds before initial ground contact, defined as the point at which the vertical ground reaction force exceeded 10N, and continued for three seconds after initial ground contact, for a total of five seconds. All flexions, adductions, and internal rotations were defined as positive angles. All extensions, abductions, and external rotations were defined as negative angles. To determine the appropriate filter for kinematic and kinetic data, a residual analysis was conducted on the ground reaction force (GRF) in a subset of the trials. Because the signal to noise ratio is dependent on the physical motion capture system and the velocities being captured and should be stable across participants, using a subset of the data is more than adequate to determine the appropriate filtering frequency. (E. Kristianslund, Krosshaug, & van den Bogert, 2012; Eirik Kristianslund, Krosshaug, & Van den Bogert, 2012; Winter, 1990). To conduct the residual analysis, raw trials from four randomly selected participants were used. Each trial was filtered in MATLAB (MathWorks, Inc., Natwick, MA) under a series of low-pass filters, yielding multiple "versions" of each GRF time series. A sum of squares was computed for each non-raw version at each time point. This can be represented as $(raw_X\text{-}filtered_{yx})^2$, where x is the frame of data and y is the filtering frequency. Once the sums of squares were computed for each frequency, residuals were obtained. For each filtering frequency, the residual is defined as the square root of the mean sum of squares across all time points (Winter, 1990). Finally, to determine the proper filtering frequency, separate plots were generated for each GRF residual. Visual inspection of the plots indicated that the optimum ratio of signal distortion to noise was approximately 10 Hz. Therefore, all kinetic and kinematic data were filtered with a 10 Hz low-pass, zero-lag, 2nd order Butterworth filter. Inverse dynamics were then computed to determine external moments for each joint (Gagnon & Gagnon, 1992), which were then normalized to each participant's height (meters) and mass (kilograms). Kinematic and kinetic data were then exported to MATLAB (MathWorks, Inc., Natwick, MA) for further reduction using custom-written script.

For reduction of MVIC and EMG data, the peak force (N) for each MVIC condition was multiplied by the moment arm length (as determined by Dempster's data and accounting for placement of the dynamometer), then divided by participant body mass, resulting in a normalized torque, $N{\cdot}m{\cdot}kg^{-1}$. These torque values were used to represent maximum torque generation. To obtain MVIC muscle activation amplitude, the EMG data were filtered in MATLAB using a band-pass 20-350 Hz, 4th order, zero-lag, Butterworth filter with full-wave rectification, and was processed using a root mean squared (RMS) algorithm with a 25-millisecond time constant. The peak RMS EMG amplitude was extracted from each trial and averaged within each condition, resulting in an averaged peak EMG amplitude for each MVIC direction that was used to normalize the EMG signal (% MVIC) obtained during the $LAND_{SL}$ and $LAND_{DL}$. Therefore, peak amplitude was defined as the averaged peak RMS signal across trials within each MVIC condition.

For conventional analyses, MATLAB script was written to extract all discrete variables from the exported data from initial ground contact (the point at which vertical ground reaction force exceeded 10N) to the end of the landing phase (the point of maximum knee flexion). Initial hip and knee angles (sagittal, frontal, and transverse) were defined as the respective joint angles at the moment of initial ground contact. Peak hip and knee angles were extracted (sagittal, frontal, and transverse planes) as the maximum respective joint angle occurring during the landing phase. Joint excursions were calculated as the peak angle minus the initial angle, in degrees. Peak external moments were obtained for the hip and knee in each cardinal plane, and were defined as the maximum normalized moment during the landing phase.

For SPM analyses, all data points of the landing phase (initial ground contact until the point of maximum knee flexion) were extracted for all biomechanical data. Custom MATLAB script was written to create time series curves of 101 data points (representing 0-100% of the landing phase), equally spaced, for each biomechanical variable and EMG signal, resulting in separate ensemble curves for each variable by sex and task.

After reduction in MATLAB, all data were exported to Excel (Microsoft Corp., Redmond, WA). Data to be used for conventional analyses were organized and transferred to SPSS (IBM Corp, Armonk, NY), where statistical analyses were conducted. Data to be used for SPM analysis remained in Excel, where it was imported into MATLAB for SPM analysis.

Statistical Approach. For the conventional analysis of discrete variables, five separate 2x2 (sex by task) RMANOVAs were used to detect differences between sex and task in peak gluteus maximus and medius muscle activation (%MVIC) and in hip and knee motion and moments for: 1) initial joint angles (º), 2) peak joint angles (º), 3) excursions (º), and 4) peak external joint moments (Nm*body weight $(N)^{-1}$*height $(m)^{-1}$). Wilk's Lambda was inspected to determine the presence of main effects. Post-hoc pairwise comparisons were used where appropriate. Statistical significance was set at 0.05 ($\alpha = 0.05$).

For SPM analysis, separate SPM 2x2 ANOVAs were used to compare the following biomechanical time series curves between sexes and tasks: knee flexion angle, knee adduction/abduction angle, knee internal/external rotation angle, hip flexion angle, hip adduction/abduction angle, hip internal/external rotation angle, knee flexion joint moment, knee adduction/abduction joint moment, knee internal/external joint moment, hip flexion joint moment, hip adduction/abduction joint moment, hip internal/external rotation joint moment, gluteus maximus muscle activation, and gluteus medius muscle activation. All SPM analyses were conducted using MATLAB script developed by Todd Pataky (Pataky, T., 2016, www.spm1d.org).

Results

General Linear Model (Conventional) Descriptive Statistics. Complete descriptive results (means ± standard deviations) for hip and knee kinematics and kinetics, and muscle activation amplitudes for each sex and task are presented in Tables 4.1-4.5.

Table 4.1. Means ± Standard Deviations (°) for Initial Joint Angles by Sex and by Task.

		Single-leg Landing	Double-leg Landing	*Total*
Knee Flexion (+)	Females	5.2±8.3	16.0±9.3	*10.6±9.8*
	Males	4.6±9.8	10.6±10.8	*7.6±10.3*
	Total	*4.9±9.0*	*13.3±10.3*	
Knee Adduction (+)	Females	1.4±6.1	3.7±6.5	*2.6±6.3*
	Males	4.2±6.0	5.0±6.3	*4.6±6.2*
	Total	*2.8±6.2*	*4.4±6.4*	
Knee Internal Rotation (+)	Females	-5.4±13.4	-2.4±12.2	*-3.9±12.8*
	Males	-5.3±10.1	-4.3±9.7	*-4.8±9.9*
	Total	*-5.4±11.8*	*-3.3±11.0*	
Hip Flexion (+)	Females	7.6±17.5	10.4±13.6	*9.0±15.6*
	Males	1.4±10.7	1.4±9.0	*1.4±9.9*
	Total	*4.5±14.8*	*5.9±12.3*	
Hip Adduction (+)	Females	-9.4±7.3	-8.5±4.9	*-8.9±6.1*
	Males	-12.0±6.0	-9.9±5.3	*-10.9±5.7*
	Total	*-10.7±6.*	*-9.2±5.1*	
Hip Internal Rotation (+)	Females	-0.7±6.7	-0.9±5.8	*-0.8±6.3*
	Males	-4.5±5.5	-2.8±4.5	*-3.7±5.0*
	Total	*-2.6±6.4*	*-1.9±5.2*	

Table 4.2. Means ± Standard Deviations (°) for Peak Joint Angles by Sex and by Task.

		Single-leg Landing	Double-leg Landing	*Total*
Knee Flexion (+)	Females	40.9±8.3	65.0±11.4	*53.0±9.9*
	Males	41.6±8.4	59.1±9.9	*50.4±9.2*
	Total	*41.3±8.3*	*62.1±11.0*	
Knee Adduction (+)	Females	6.0±6.2	7.5±9.8	*6.8±8.0*
	Males	7.9±6.4	8.1±6.0	*8.0±6.2*
	Total	*6.9±6.3*	*7.8±8.1*	
Knee Abduction (-)	Females	-3.1±5.7	-5.5±7.3	*-4.3±6.5*
	Males	0.6±6.5	-0.9±6.7	*-0.2±6.6*
	Total	*-1.2*±6.4	*-3.2*±7.4	
Knee Internal Rotation (+)	Females	12.0±12.7	12.6±13.2	*12.3±13.0*
	Males	12.6±9.4	13.8±9.5	*13.2±9.5*
	Total	*12.3±11.1*	*13.2±11.4*	
Knee External Rotation (-)	Females	-8.1±11.3	-7.6±11.7	*-7.9±11.5*
	Males	-7.5±9.6	0.9±5.4	*-7.3±7.5*
	Total	-7.8±10.4	-7.3±10.5	
Hip Flexion (+)	Females	19.8±16.6	27.1±13.4	*23.5±15.0*
	Males	13.2±11.3	17.8±10.7	*15.5±11.0*
	Total	*16.5±14.5*	*22.4±12.9*	
Hip Adduction (+)	Females	0.9±5.4	-3.8±5.6	*-1.5±5.5*
	Males	-1.3±7.8	-5.9±7.2	*-3.6±7.5*
	Total	-0.2±6.8	-4.8±6.5	
Hip Abduction (-)	Females	-12.3±5.9	-11.2±6.5	*-11.7±6.2*
	Males	-13.7±5.8	-11.9±6.0	*-12.8±5.9*
	Total	*-13.0±5.9*	*-11.5±6.2*	
Hip Internal Rotation (+)	Females	1.1±6.2	1.0±4.9	*1.1±5.6*
	Males	-1.0±5.0	-0.6±4.2	*-0.8±4.6*
	Total	*0.0±5.7*	*0.2±4.6*	
Hip External Rotation (-)	Females	-6.2±6.4	-8.1±6.1	*-7.1±6.3*
	Males	-7.5±4.9	-8.1±4.3	*-7.8±4.6*
	Total	-6.8±5.7	-8.1±5.2	

Table 4.3. Means ± Standard Deviations (°) for Joint Excursions by Sex and by Task.

		Single-leg Landing	Double-leg Landing	*Total*
Knee Flexion (+)	Females	35.7±10.4	49.0±15.4	*42.4±12.9*
	Males	37.0±11.2	48.5±13.1	*42.8±12.2*
	Total	*36.4±10.8*	*48.8±14.2*	
Knee Adduction (+)	Females	4.6±4.6	3.6±6.0	*4.2±5.3*
	Males	3.6±2.8	3.1±3.1	*3.4±3.0*
	Total	*4.1±3.8*	*3.4±4.8*	
Knee Abduction (-)	Females	-4.4±3.5	-9.3±6.3	*-6.9±4.9*
	Males	-3.7±2.8	-6.0±4.7	*-4.8±3.8*
	Total	*-4.0±3.2*	*-7.6±5.8*	
Knee Internal Rotation (+)	Females	17.4±8.1	15.0±8.3	*16.2±8.2*
	Males	18.0±6.4	18.1±8.1	*18.0±7.3*
	Total	*17.7±7.3*	*16.5±8.3*	
Knee External Rotation (-)	Females	-2.7±3.1	-5.2±4.9	*-4.0±4.0*
	Males	-2.1±2.1	-2.8±2.6	*-2.4±2.4*
	Total	*-2.4±2.7*	*-4.0±4.1*	
Hip Flexion (+)	Females	12.2±11.2	16.7±13.3	*14.4±12.3*
	Males	11.8±7.2	16.4±8.9	*14.1±14.1*
	Total	*12.0±9.4*	*16.5±11.2*	
Hip Adduction (+)	Females	10.3±5.3	4.7±3.2	*7.5±4.3*
	Males	10.7±5.0	4.0±3.2	*7.3±4.1*
	Total	*10.5±5.1*	*4.3±3.2*	
Hip Abduction (-)	Females	-2.9±3.0	-2.7±4.9	*-2.8±4.0*
	Males	-1.7±1.8	-2.0±2.1	*-1.8±2.0*
	Total	*-2.3±2.5*	*-2.3±3.8*	
Hip Internal Rotation (+)	Females	1.7±2.3	2.0±2.5	*1.9±2.4*
	Males	3.5±2.9	2.2±2.2	*2.8±2.6*
	Total	*2.6±2.7*	*2.1±2.4*	
Hip External Rotation (-)	Females	-5.5±2.9	-7.2±3.9	*-6.3±3.4*
	Males	-3.0±2.3	-5.3±3.4	*-4.1±2.9*
	Total	*-4.2±2.9*	*-6.2±3.7*	

Table 4.4. Means ± Standard Deviations (Nm/(N*m)) for Peak External Joint Moments by Sex and by Task.

		Single-leg Landing	Double-leg Landing	*Total*
Knee Flexion (+)	Females	.110±.058	.098±.035	*.104±.047*
	Males	.094±.056	.099±.060	*.097±.058*
	Total	*.102±.056*	*.099±.049*	
Knee Adduction (+)	Females	.096±.078	.046±.040	*.069±.059*
	Males	.097±.043	.054±.040	*.075±.042*
	Total	*.096±.063*	*.049±.040*	
Knee Abduction (-)	Females	-.018±.031	-.019±.024	*-.019±.028*
	Males	-.014±.026	-.015±.024	*-.014±.025*
	Total	*-.016±.029*	*-.017±.024*	
Knee Internal Rotation (+)	Females	.020±.015	.014±.016	*.017±.016*
	Males	.025±.016	.016±.016	*.021±.016*
	Total	*.023±.015*	*.015±.016*	
Knee External Rotation (-)	Females	-.003±.004	-.003±.003	*-.003±.003*
	Males	-.002±.003	-.003±.003	*-.002±.003*
	Total	*-.002±.003*	*-.003±.003*	
Hip Flexion (+)	Females	.116±.113	.079±.062	*.097±.088*
	Males	.099±.055	.122±.173	*.110±.114*
	Total	*.107±.089*	*.100±.131*	
Hip Adduction (+)	Females	.162±.070	.084±.138	*.123±.104*
	Males	.146±.051	.077±.061	*.112±.056*
	Total	*.154±.061*	*.081±.106*	
Hip Abduction (-)	Females	-.030±.067	-.042±.117	*-.036±.092*
	Males	-.015±.041	-.030±.046	*-.023±.044*
	Total	*-.022±.055*	*-.036±.089*	
Hip Internal Rotation (+)	Females	.012±.017	.012±.025	*.012±.021*
	Males	.009±.016	.013±.024	*.011±.020*
	Total	*.010±.017*	*.013±.025*	
Hip External Rotation (-)	Females	-.023±.014	-.014±.017	*-.019±.016*
	Males	-.024±.016	-.019±.022	*-.021±.019*
	Total	*-.024±.015*	*-.016±.019*	

Table 4.5. Means ± Standard Deviations for Gluteal Activation (%MVIC) by Sex and by Task.

		Single-Leg Landing	Double-Leg Landing	*Total*
GMax Peak Activation	Females	37.2±31.4	30.3±31.9	*33.7±31.7*
	Males	31.9±29.2	31.9±39.1	*31.5±34.2*
	Total	*34.5±30.3*	*30.7±35.4*	
GMed Peak Activation	Females	42.2±23.7	29.4±19.9	*35.8±21.8*
	Males	80.6±114.6	49.7±46.7	*65.2±161.3*
	Total	*61.4±84.5*	*39.6±37.1*	

Omnibus MANOVA Results. Initial hip and knee joint angles differed by sex (λ=.82, p=.01), task (λ=.36, p<.001), and sex by task (λ=.80, p=.01). Likewise, peak hip and knee joint angles differed by sex (λ=.76, p=.01), task (λ=.08, p<.001), and sex by task (λ=.80, p=.05), and joint excursions differed by sex (λ=.75, p=.01), task (λ=.11, p<.001), and sex by task (λ=.75, p=.01).

Peak external joint moments differed by task (λ=.10, p<.001). There were no differences between sex (λ=.86, p=.26) or sex by task (λ=.86, p=.23).

Gluteal muscle activation amplitudes (%MVIC) differed by sex (λ=.90, p=.01) and task (λ=.92, p=.03). There were no sex by task interactions (λ=.89, p=.29).

Univariate Results. Univariate results will be presented by joint and plane along with the SPM analyses. In this way, results obtained from both GLM and SPM analyses can be considered together, making for greater cohesion and a more holistic description and interpretation of movement patterns.

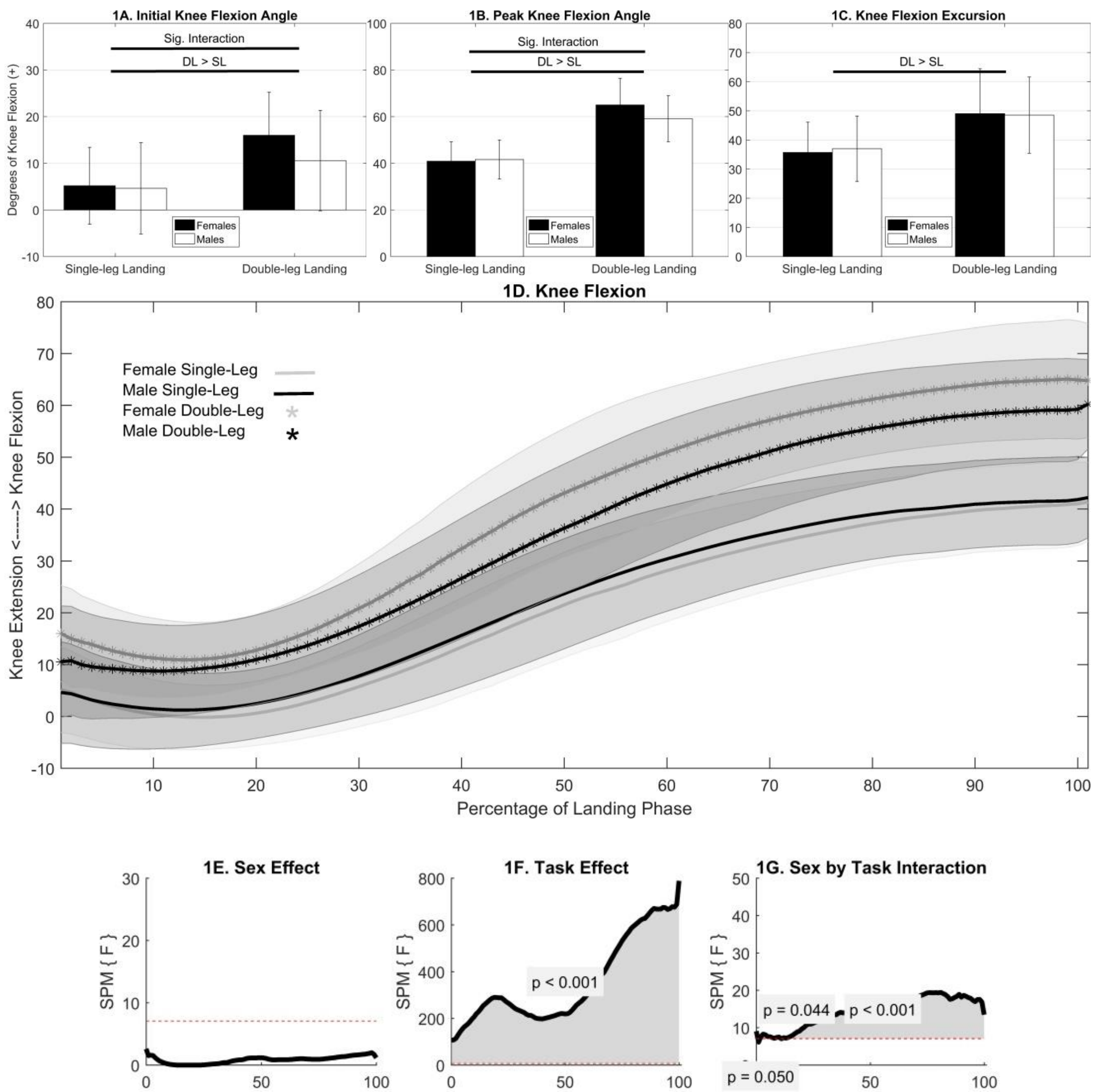
***Statistical Parametric Mapping Analysis*.** To properly interpret results from an SPM analysis, two sets of graphs are necessary. The first set is descriptive. Examples of a descriptive set can be seen in Figure 4.1d, in which knee flexion curves are depicted for males and females in each landing task. These curves are time-normalized to 100% of the landing phase, as indicated along each x-axis. The second set of graphs depicts the inferential analysis, examples of which are presented in Figures 4.1e-g. For each inferential set of graphs, sex effects, task effects, and interaction effects are depicted separately. The dotted line in each graph represents the critical F-statistic, above which lies statistical significance. Similar to the descriptive graphs, % landing phase is along the x-axis of the inferential graphs. In this way, the *moment(s)* in time at which differences occur during the landing phase can be determined, while the specific directional differences can be observed and interpreted from the descriptive graphs.

Knee Flexion Kinematics. Univariate analysis revealed a sex by task interaction for initial (p=.003) and peak (p<.001) knee flexion angles and a main effect for task in knee flexion excursion (p<.001). Compared to males, females landed in greater initial (Δ=5.4°, d=.54) and peak (Δ=5.9°, d=.55) knee flexion during $LAND_{DL}$, but were similar to males in initial (Δ=0.6°, d=.07) and peak knee flexion (Δ= -0.7°, d=.08) during $LAND_{SL}$ (Figures 4.1a and 4.1b). In both sexes, sagittal plane knee excursions were 12.4° smaller during $LAND_{SL}$ than $LAND_{DL}$ (d=.98) (Figure 4.3c).

SPM analysis also revealed a sex by task interaction in sagittal plane knee kinematics ($F_{crit(1,88)}$=7.09). Females displayed less knee flexion throughout the landing phase of $LAND_{SL}$, but greater knee flexion than males throughout $LAND_{DL}$. This interaction effect was present in the following supra-threshold clusters, described as percentages of the landing phase: 0-1% (p=.05), 2-10% (p=.04), and 11-100% (p<.001) (Figure 4.1g).

SPM confirmed that the observed interactions in GLM initial and peak knee flexion angle analysis were not isolated to these discrete time points, but were present throughout the landing phase.

Figure 4.1. Knee Flexion: Univariate Results (1a-c) and SPM Descriptive and Inferential Results (1d-g).

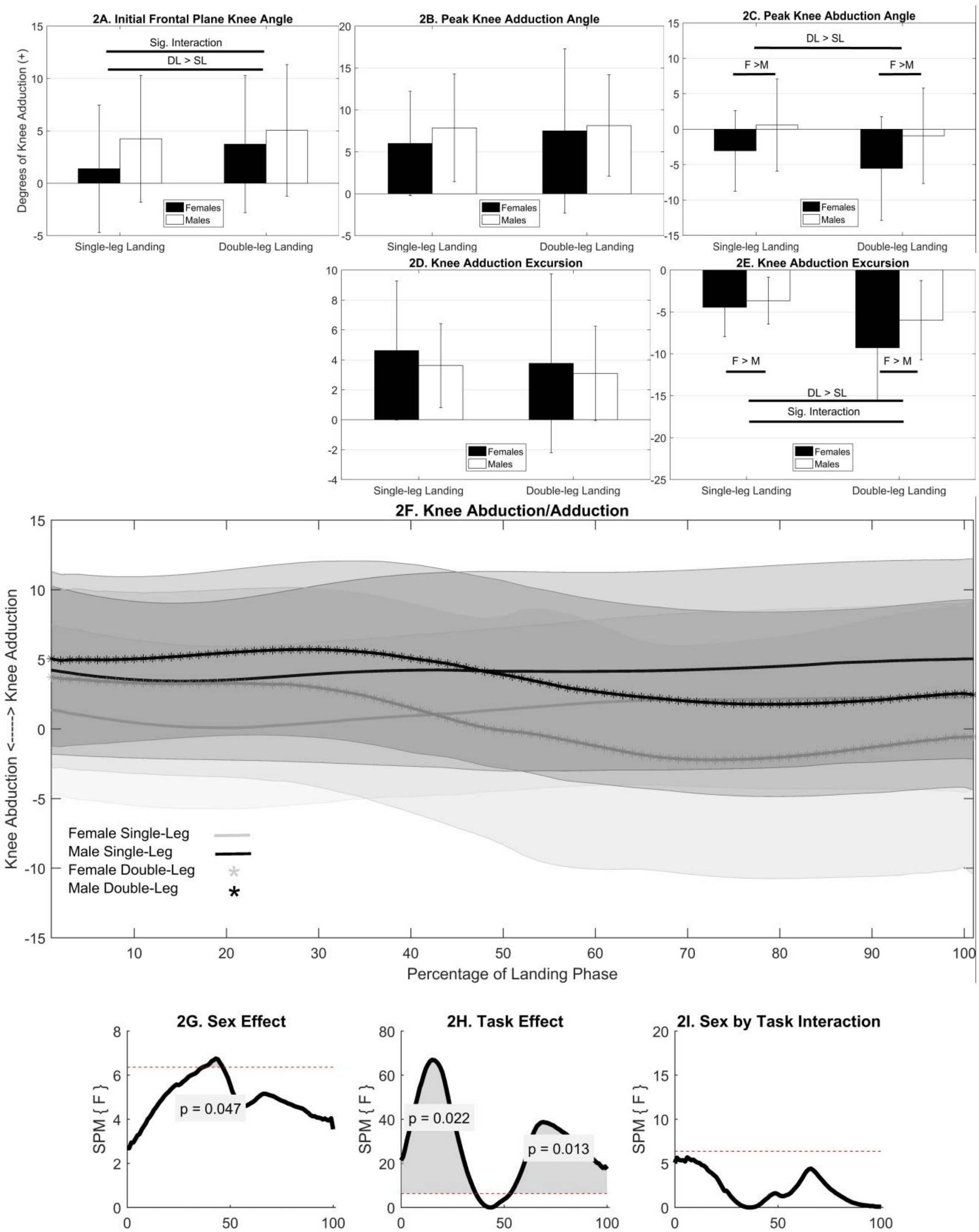


Knee Frontal Plane Kinematics. Univariate analysis identified sex by task interactions for frontal plane knee angles at initial contact (p=.03) and total excursions (p=.01), and main effects for sex (p=.002) and task (p<.001) for peak angles. Females made ground contact in greater knee abduction and went through greater knee abduction excursions than males in both tasks. However, the sex difference during $LAND_{DL}$ was smaller for initial contact angles (Δ=1.5°; $LAND_{SL}$: d=.46, $LAND_{DL}$: d=.20) (Figure 4.2a) and larger for total excursion (Δ=2.6°; $LAND_{SL}$: d=.22, $LAND_{DL}$: d=.59) (Figure 4.2e) than during $LAND_{SL}$. Females exhibited 4.1° greater peak knee abduction regardless of task (d=.63) (Figure 4.2c), and $LAND_{DL}$ elicited 2.0° greater peak knee abduction angles than $LAND_{SL}$ in both sexes (d=.29) (Figure 4.2c).

SPM analysis revealed sex differences and task differences in frontal plane knee movement ($F_{crit(1,88)}$=6.37), but no sex by task interaction. Regardless of task, females were more abducted than males, but this was constrained to 37-46% of the landing phase (p=.05) (Figure 4.2g). Task differences were located in two distinct supra-threshold clusters. $LAND_{SL}$ resulted in more abducted knees from 0-36% than did the corresponding section of $LAND_{DL}$ (p=.02). Conversely, from 54-100%, participants displayed greater knee abduction during $LAND_{DL}$ as compared to $LAND_{SL}$ (p=.01) (Figure 4.2h).

SPM did not confirm the presence of interactions in initial angles or total frontal plane knee excursions, but did agree with GLM on the presence of main effects for sex and task, and further served to specify that the timing of these differences occurred earlier during $LAND_{SL}$ than in $LAND_{DL}$.

Figure 4.2. Knee Adduction/Abduction: Univariate Results (2a-e) and SPM Descriptive and Inferential Results (2f-i).

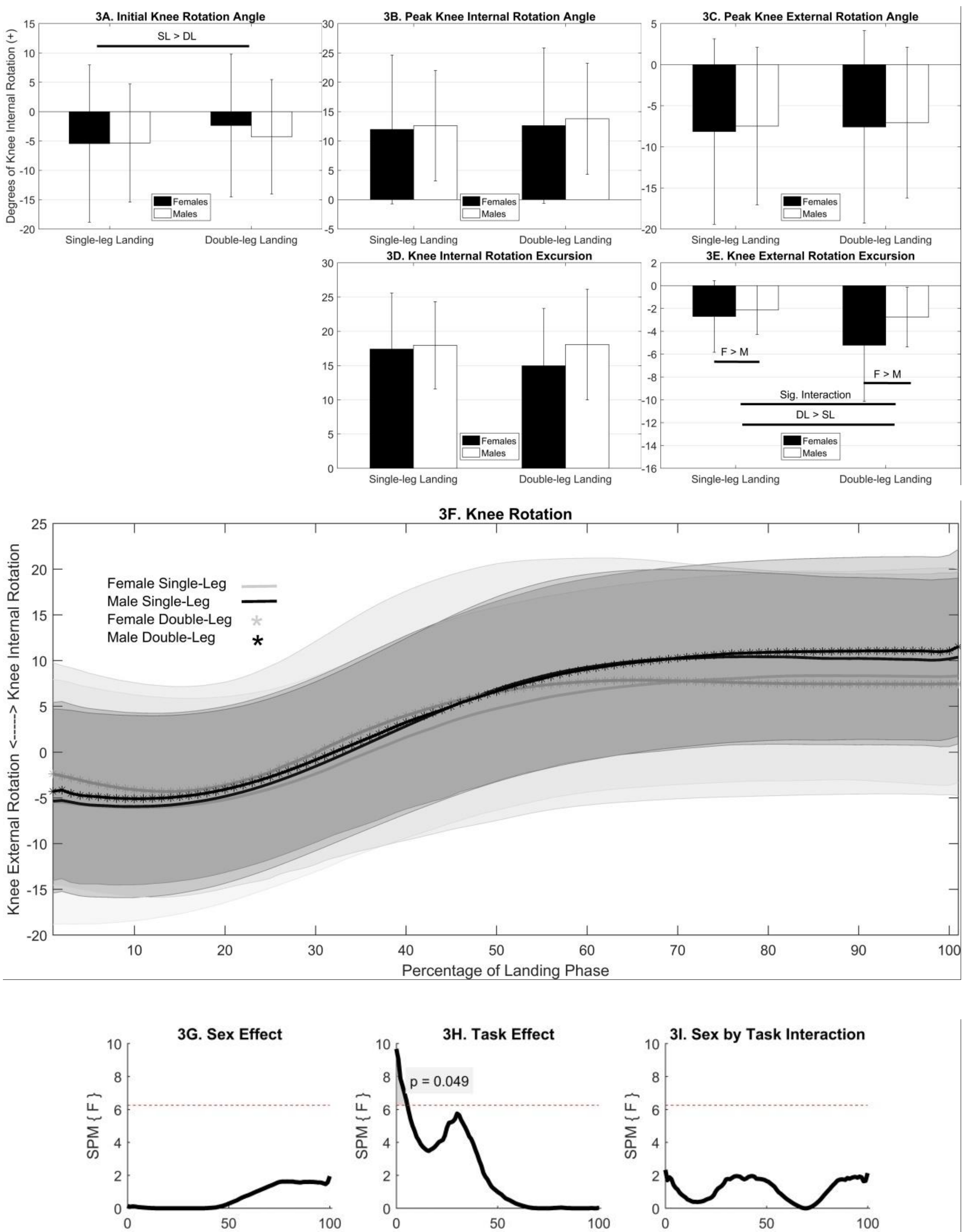


Knee Transverse Plane Kinematics. Univariate analysis revealed a sex by task interaction for knee external rotation excursions (*p*=.03) and a main effect for task for initial contact angles (*p*=.003). Compared to males, females moved through 2.4° more knee external rotation during the $LAND_{DL}$ (*d*=.61), but were similar to males during $LAND_{SL}$ (0.6° greater, *d*=.23) (Figure 4.3e). Regardless of sex, $LAND_{SL}$ elicited 2.1° greater knee external rotation at initial ground contact than did $LAND_{DL}$ (*d*=.18) (Figure 4.3a).

SPM analysis demonstrated that males and females displayed greater knee internal rotation during 0-5% of $LAND_{DL}$ when compared to $LAND_{SL}$ ($F_{crit(1,88)}$=6.26, *p*=.05) (Figure 4.3h).

SPM confirmed the presence of a task effect at initial ground contact, limiting this effect to the first 5% of the landing phase. SPM did not identify the interaction in joint excursions revealed by GLM when all times points were considered.

Figure 4.3. Knee Rotation: Univariate Results (3a-e) and SPM Descriptive and Inferential Results (3f-i).

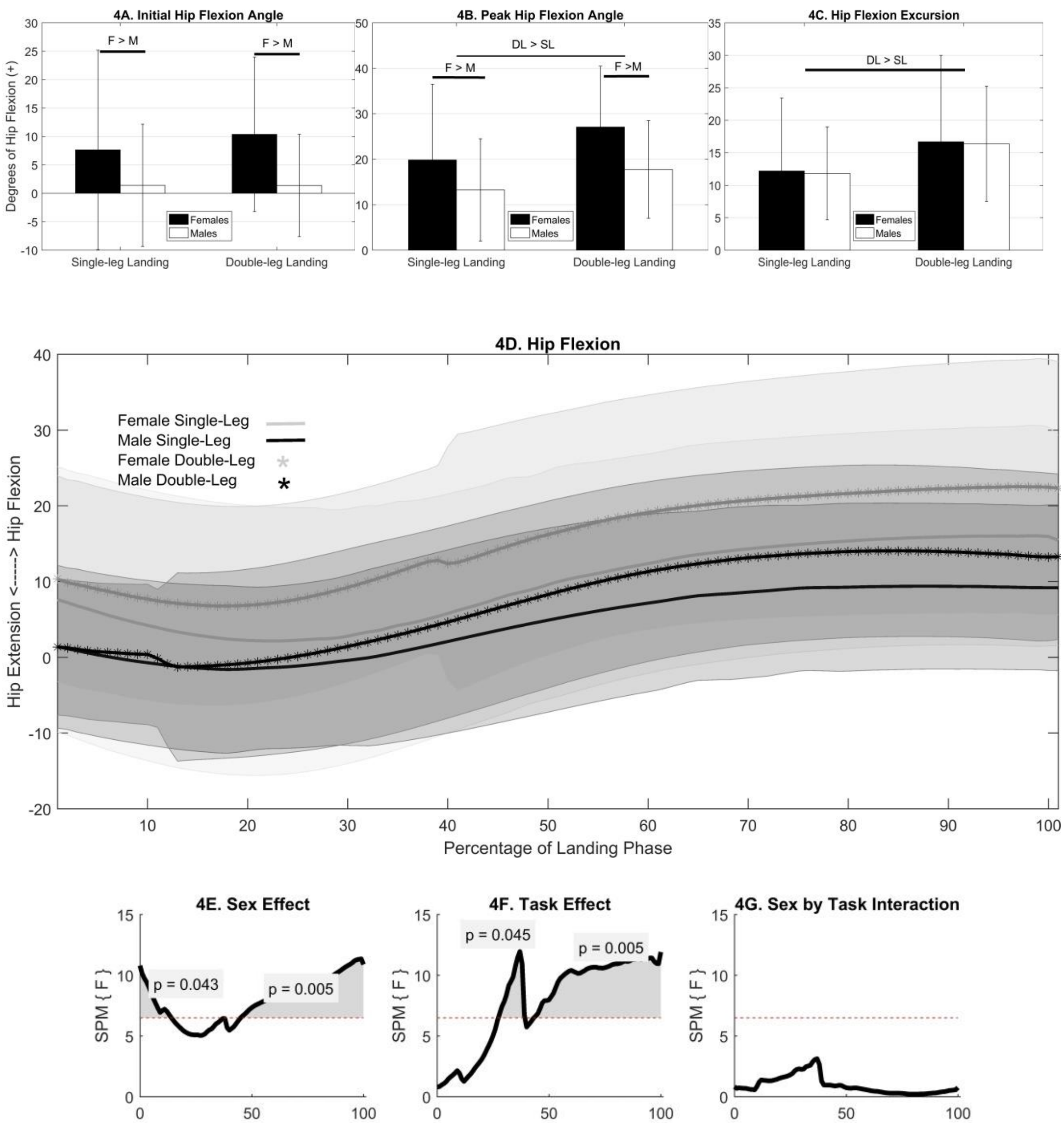


Hip Flexion Kinematics. Univariate analysis revealed main effects for sex in initial (p=.001) and peak hip flexion (p=.001) and main effects for task in peak hip flexion (p<.001) and total excursions (p<.001). Females made ground contact with 7.6° more hip flexion (d=.58) and displayed 8.0° greater peak hip flexion angles (d=.61) than males (Figures 4.4a and 4.4b). Subjects performed $LAND_{DL}$ by going through 4.5° (d=.44) greater motion resulting in 5.9° greater peak hip angles (d=.43) than $LAND_{SL}$ (Figures 4.4b and 4.4c). There were no sex by task interactions.

SPM analysis also revealed main effects for sex and task ($F_{crit(1,88)}$=6.50), and indicated that the greater hip flexion exhibited in females occurred from 0-14% (p=.04) and from 45-100% (p=.01) of the landing phase (Figure 4.4e). Greater flexion in $LAND_{DL}$ vs $LAND_{SL}$ occurred from 28-39% (p=.05) and 44-100% (p=.01) (Figure 4.4f). There was no sex by task interaction.

SPM confirmed the GLM sex effect at initial ground contact, and clarified that the greater peak hip flexion in females is largely occurring in the latter half of the landing phase. SPM also further elucidated the GLM task effects for peak hip flexion and excursions, indicating that differences begin to develop at approximately 30% into the landing phase, and continue to grow for the remainder of the task.

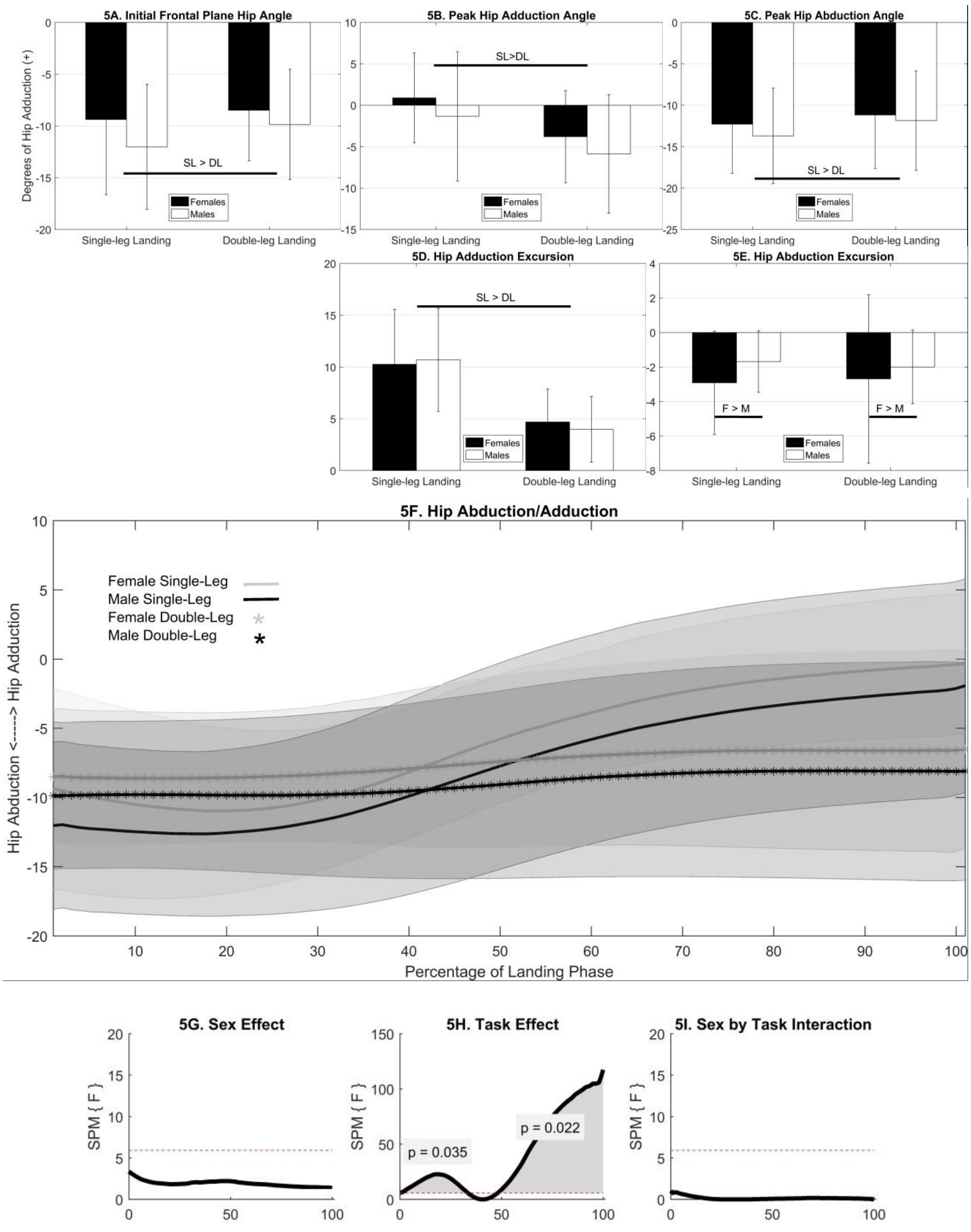
Figure 4.4. Hip Flexion: Univariate Results (4a-c) and SPM Descriptive and Inferential Results (4d-g).



Hip Frontal Plane Kinematics. Univariate analysis revealed a main effect for sex for hip abduction excursion (p=.05) and main effects for task for initial angle (p=.02), peak hip adduction (p<.001), peak hip abduction (p=.01), and hip adduction excursion (p<.001). Females went through 1.0° more hip abduction excursion than males in both tasks (d=.32) (Figure 4.5e). Compared to $LAND_{DL}$, subjects displayed 1.5° greater hip abduction at initial contact (d=.25) (Figure 4.4a), 4.6° greater peak hip adduction (d=.69) (Figure 4.4b), 1.5° greater peak hip abduction (d=.25) (Figure 4.4c), and 6.2° greater hip adduction excursion (d=1.46) (Figure 4.4d) during $LAND_{SL.}$

SPM did not confirm the GLM main effect for sex, but did complement findings from GLM regarding the timing of peak adduction and abduction angles. Specifically, SPM analysis identified two distinct supra-threshold clusters ($F_{crit(1,88)}$=5.92), with greater hip abduction observed from 0-33% (p=.04), then greater hip adduction observed from 49-100% (p=.02) (Figure 4.5h) of the $LAND_{SL}$ than during the corresponding sections of the $LAND_{DL}$. Together, this would explain the greater hip adduction excursion observed in the GLM.

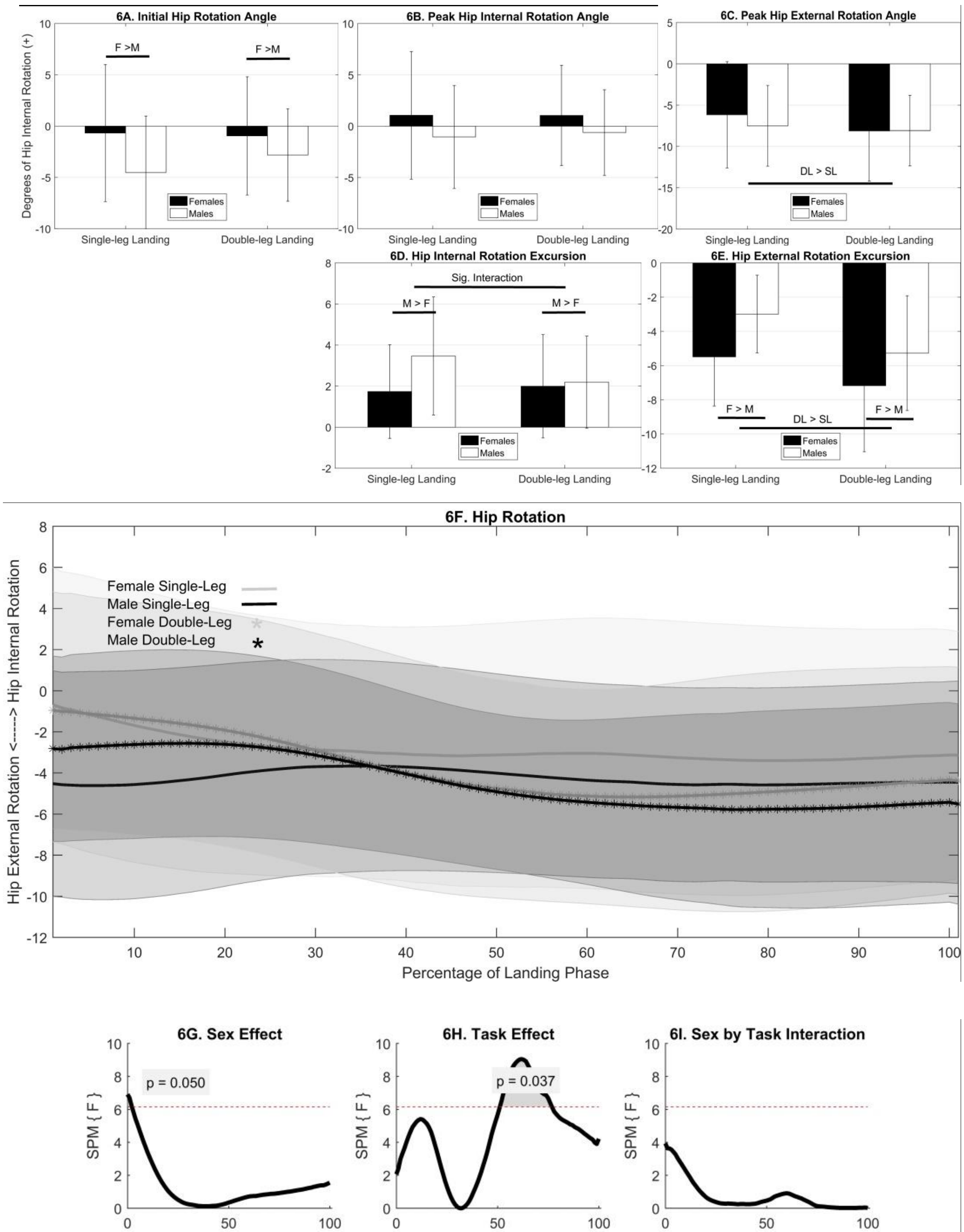
Figure 4.5. Hip Adduction/Abduction: Univariate Results (5a-e) and SPM Descriptive and Inferential Results (5f-i).



Hip Transverse Plane Kinematics. Univariate analysis revealed a sex by task interaction for internal rotation excursion (p=.03), main effects for sex in initial angle (p=.01) and external rotation excursion (p<.001), and main effects for task in peak external rotation (p=.02) and external rotation excursion (p<.001). Females made ground contact with 2.9° greater internal rotation (d=.51), then moved through 2.2° more external rotation (d=.70) than males (Figures 4.6a and 4.6e). For internal rotation excursion, females moved through 1.8° less internal rotation during $LAND_{SL}$ (d=.69), but were more similar to males in $LAND_{DL}$ (0.2° less; d=.08) (Figure 4.6d). $LAND_{DL}$ elicited 1.3° greater peak external rotation angles (d=.24) and 2.0° greater external rotation excursions than $LAND_{SL}$ (d=.60) (Figures 4.6c and 4.6e).

SPM analysis revealed main effects for sex and task ($F_{crit(1,88)}$=6.15), but did not confirm the sex by task interaction. Females only displayed greater hip internal rotation than males from 0-2% of the landing phase (p=.05) (Figure 4.6g), which supports the GLM results for sex differences in initial angles, but not total excursions. For task, the greater peak hip external rotation during $LAND_{DL}$ was limited to 52-76% of the landing phase when compared to the corresponding section of $LAND_{SL}$ (p=.04) (Figure 4.6h).

Figure 4.6. Hip Rotation: Univariate Results (5a-e) and SPM Descriptive and Inferential Results (5f-i).

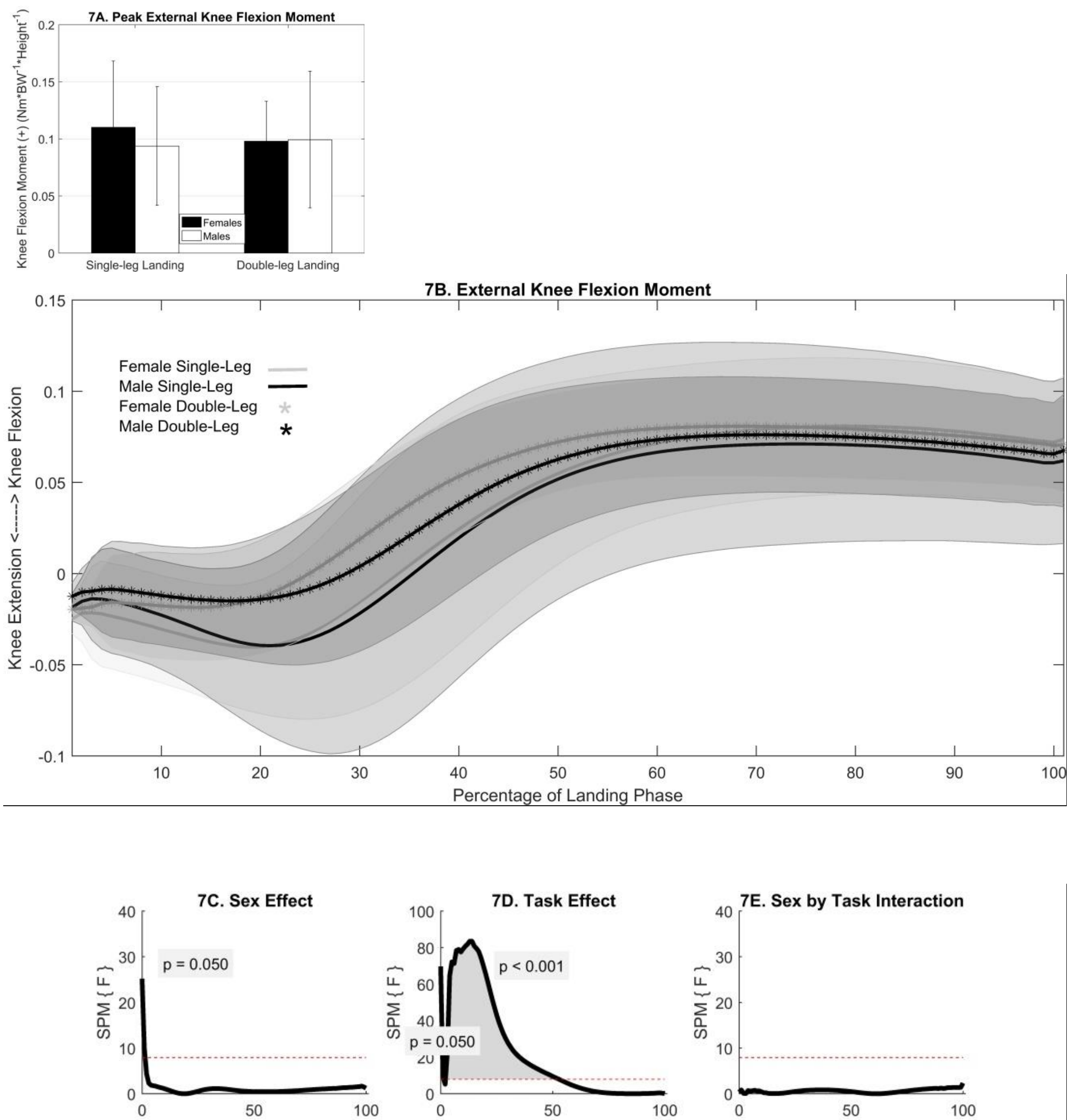


Knee Flexion Kinetics. Univariate analysis revealed no sex or task main effects nor a sex by task interaction for peak knee flexion moment (Figure 4.6a).

SPM analysis identified both sex and task main effects ($F_{crit(1,88)}$=7.94). From 0-1% of the landing phase, males displayed a greater knee flexion moment (p=.05) than females (Figure 4.7c). $LAND_{DL}$ elicited larger knee flexion moments from 0-1% (p=.05) and from 3-52% (p<.001) than did $LAND_{SL}$ (Figure 4.7d).

SPM analysis identified sex and task differences in knee flexion joint moments, both at initial contact, and throughout the landing phase, whereas GLM did not identify differences in peak joint moment. SPM identified differences in the first half of the landing phase, while the peak joint moment occurred in the latter half of the landing task.

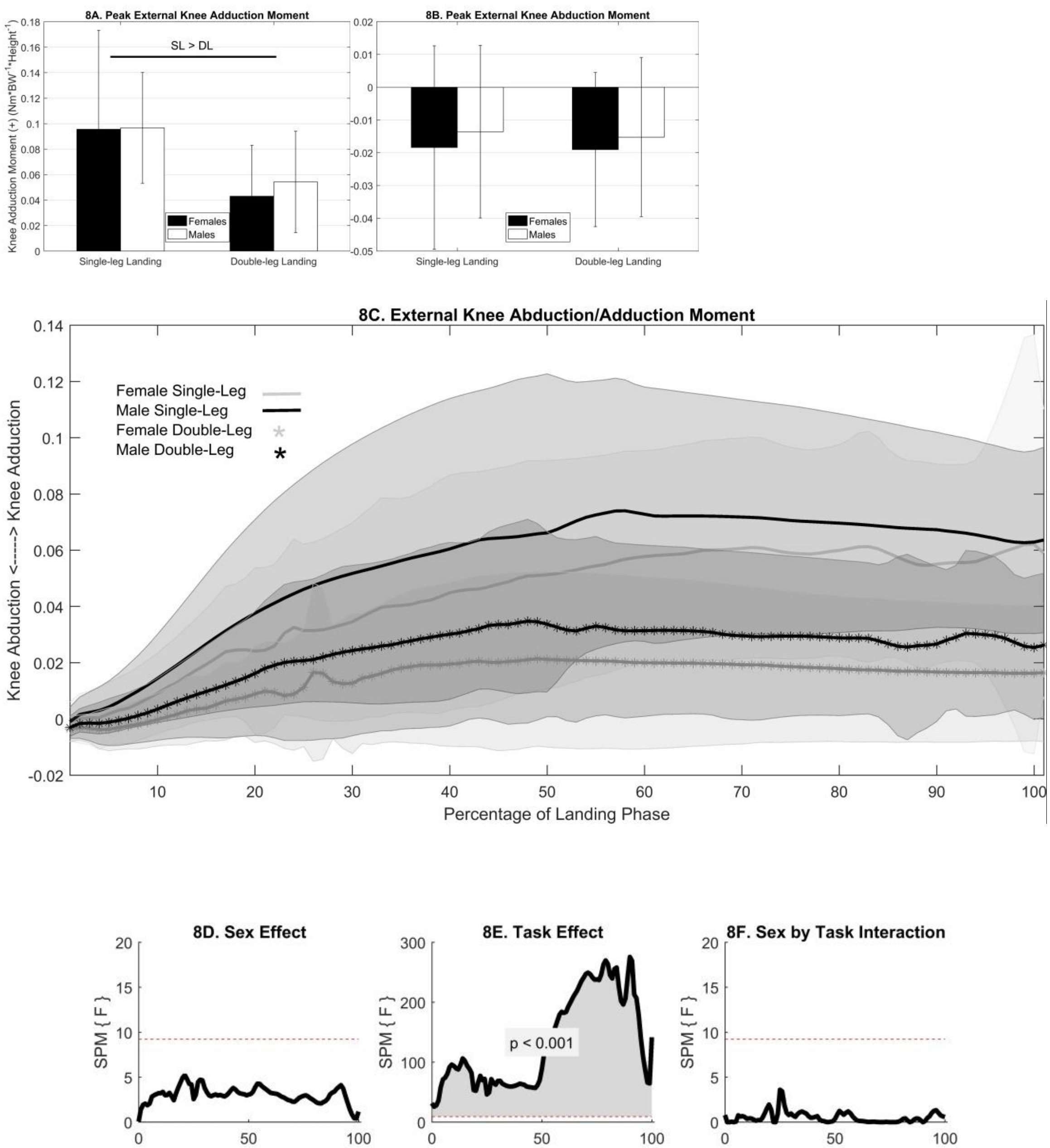
Figure 4.7. Knee Flexion Moment: Univariate Results (6a) and SPM Descriptive and Inferential Results (6b-e).



Knee Frontal Plane Kinetics. Univariate analysis of GLM results revealed a main effect for task (p<.001). $LAND_{SL}$ elicited .047 Nm/N*m greater knee moment than did $LAND_{DL}$ (d=.89) (Figure 4.8a).

SPM analysis also revealed a main effect for task ($F_{crit(1,88)}$=9.23), and further clarifies that these larger knee adduction moments were observed throughout the entire landing phase in $LAND_{SL}$ vs. $LAND_{DL}$ (0-100%; p<.001) (Figure 4.8e).

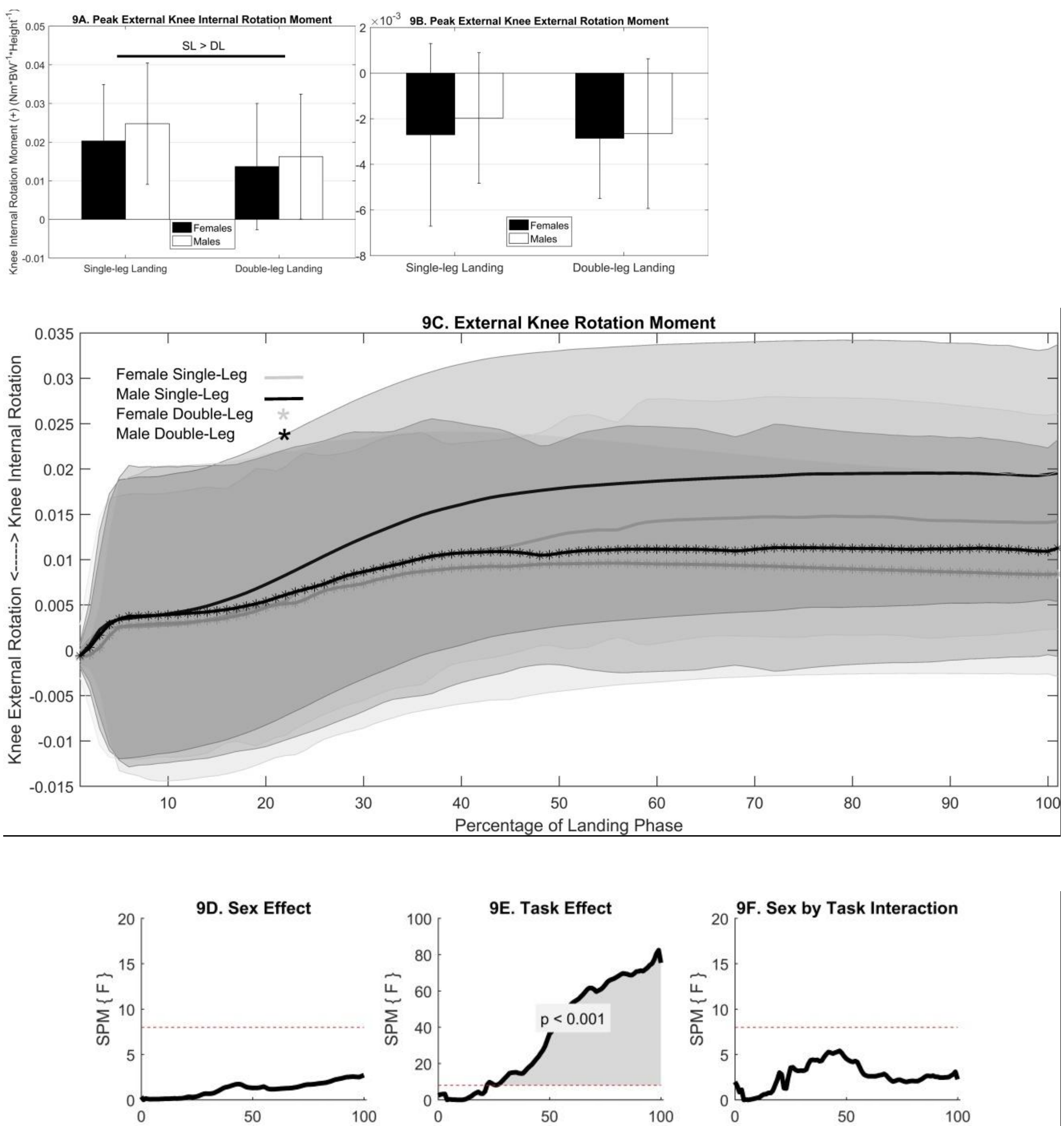
Figure 4.8. Knee Adduction/Abduction Moment: Univariate Results (8a-b) and SPM Descriptive and Inferential Results (8c-f).



Knee Transverse Plane Kinetics. Univariate analysis revealed a main effect for task in peak internal rotation joint moment (p<.001). $LAND_{SL}$ elicited .008 greater normalized Nm than did $LAND_{DL}$ (d=.52) (Figure 4.9a).

SPM analysis also revealed a main effect for task ($F_{crit(1,88)}$=8.02), and clarified that these moments were greater in $LAND_{SL}$ vs $LAND_{DL}$ through much of the landing phase (22-100%; p<.001) (Figure 4.9e).

Figure 4.9. Knee Rotation Moment: Univariate Results (9a-b) and SPM Descriptive and Inferential Results (9c-f).

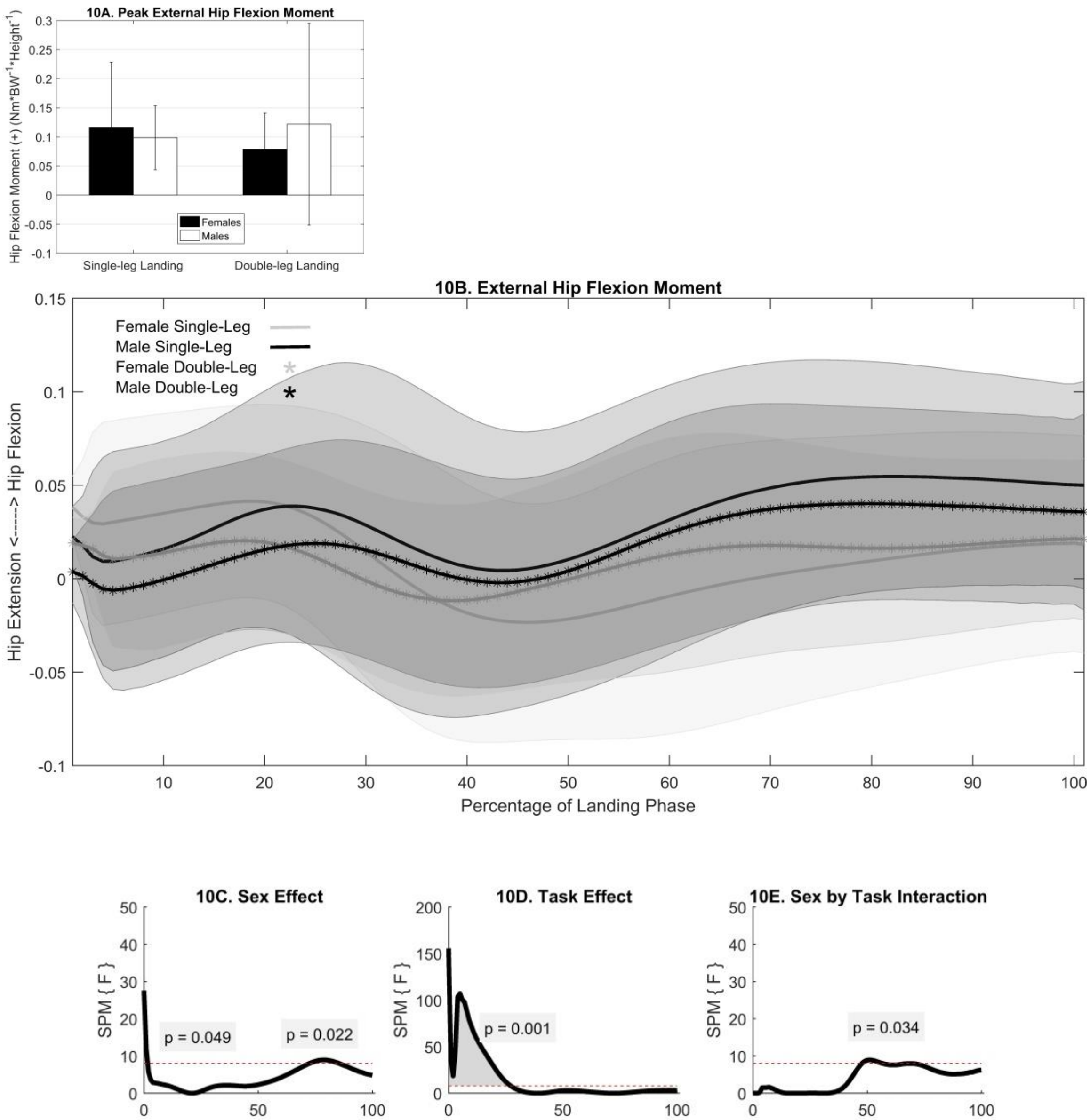


Hip Flexion Kinetics. Univariate analysis of GLM results revealed no sex by task interactions or main effects for sex or task (Figure 4.10a).

SPM analysis revealed a sex by task interaction, main effects for sex, and a main effect for task ($F_{crit(1,88)}$=8.03). Females exhibited greater hip flexion moments at initial contact (0-1%; p=.05) but males displayed greater hip flexion moments from 73-85% of the landing phase (p=.02) (Figure 4.10c). Males also exhibited greater hip flexion moments from 48-56%, but this effect was more pronounced during $LAND_{SL}$ than $LAND_{DL}$ (p=.03) (Figure 4.10e). Hip flexion moments were greater during the first quarter (0-27%) of the $LAND_{SL}$ compared to $LAND_{DL}$ (p=.001) (Figure 4.10d).

SPM analysis identified sex, task, and sex by task differences in hip flexion joint moments, both at initial contact, and throughout the landing phase, whereas GLM did not identify any main effects or interactions pertaining to peak joint moment. SPM identified differences primarily in the first half of the landing phase, while GLM examined the peak joint moment located in the latter half of the landing task.

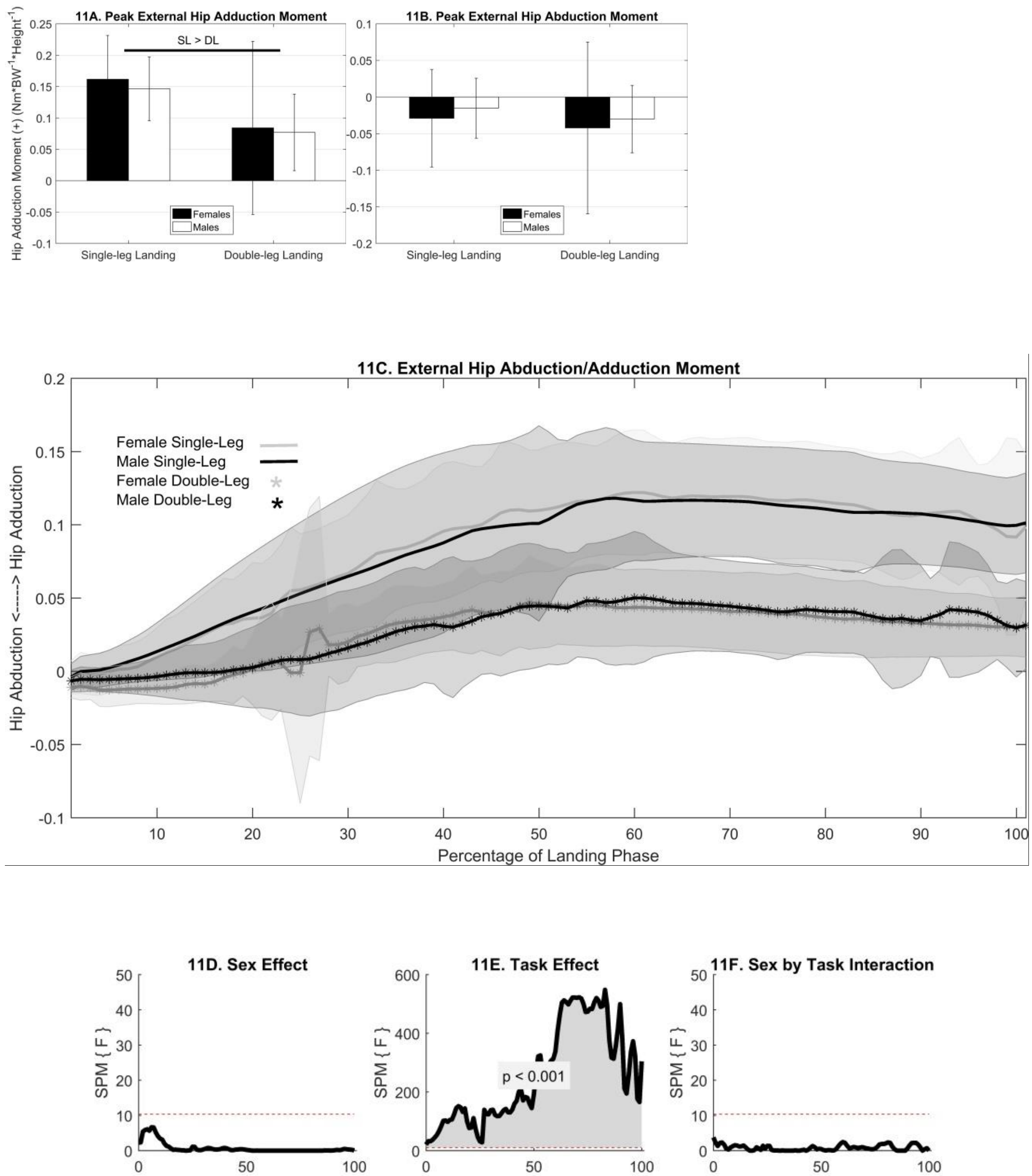
Figure 4.10. Hip Flexion Moment: Univariate Results (10a) and SPM Descriptive and Inferential Results (10b-e).



Hip Frontal Plane Kinetics. Univariate analysis revealed a main effect for task in peak adduction joint moment (p<.001) (Figure 4.11a) with .073 (Nm/N*m) higher moments observed in LAND_{SL} versus LAND_{DL} (d=.84).

SPM analysis also revealed a main effect for task ($F_{crit(1,88)}$=10.36), confirming the greater hip adduction moments during LAND_{SL} occurred throughout the landing phase (0-100%) when compared to the LAND_{DL} (p<.001) (Figure 4.11e).

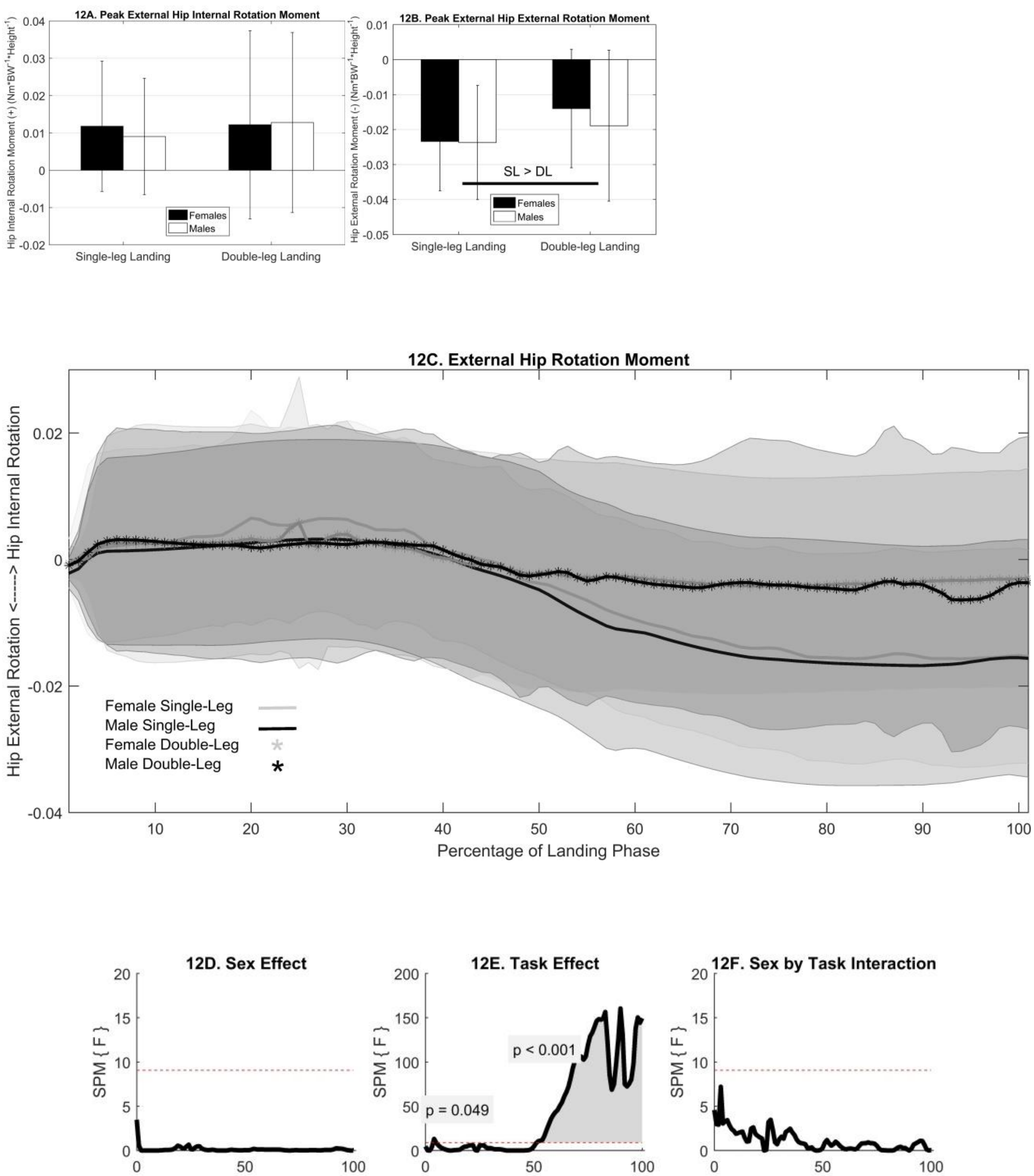
Figure 4.11. Hip Adduction/Abduction Moment: Univariate Results (11a-b) and SPM Descriptive and Inferential Results (11c-f).



Hip Transverse Plane Kinetics. Univariate analysis revealed a main effect for task in external rotation joint moment (p=.002), where moments were .008 (Nm/N*m) greater in $LAND_{SL}$ vs. $LAND_{DL}$ (d=.47) (Figure 4.12b).

SPM analysis also revealed main effects for task ($F_{crit(1,88)}$=9.08), and clarified that the greater moments during $LAND_{SL}$ were constrained to 4% (p=.05) and from 52-100% (p<.001) of the landing phase when compared with the corresponding section of $LAND_{DL}$ (Figure 4.12e).

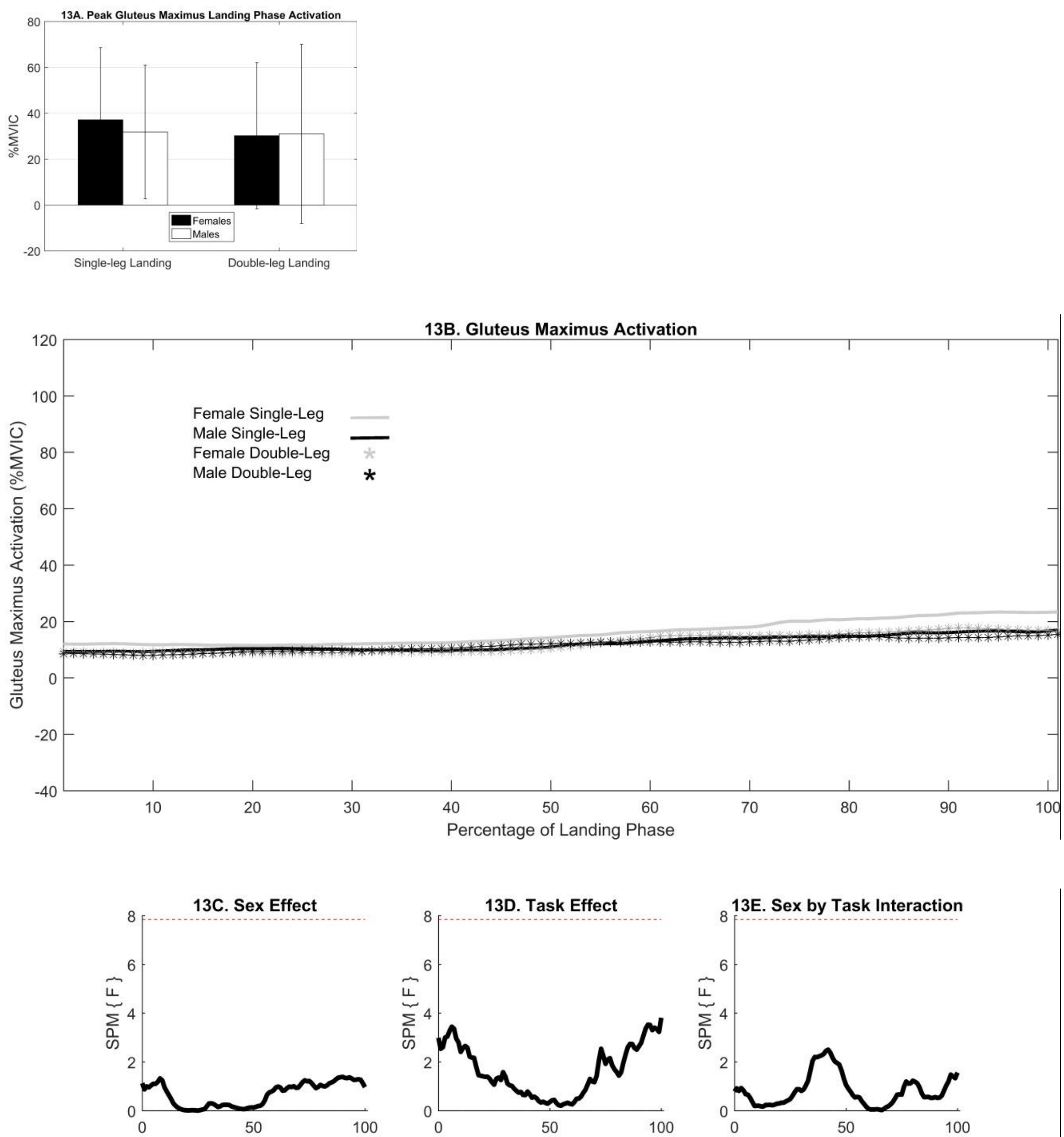
Figure 4.12. Hip Rotation Moment: Univariate Results (12a-b) and SPM Descriptive and Inferential Results (12c-f).



Maximal Voluntary Isometric Contractions. Normalized to body mass, males generated greater maximal torque values than females for hip extension (1.04±.26 v. 0.86±.21 Nm/kg, p=.001) and hip abduction (1.49±.29 v. 1.16±.24 Nm/kg, p<.001).

Gluteus Maximus Activation. Neither GLM nor SPM analyses revealed sex by task interactions or sex or task main effects for gluteus maximus activation (Figure 13a-f).

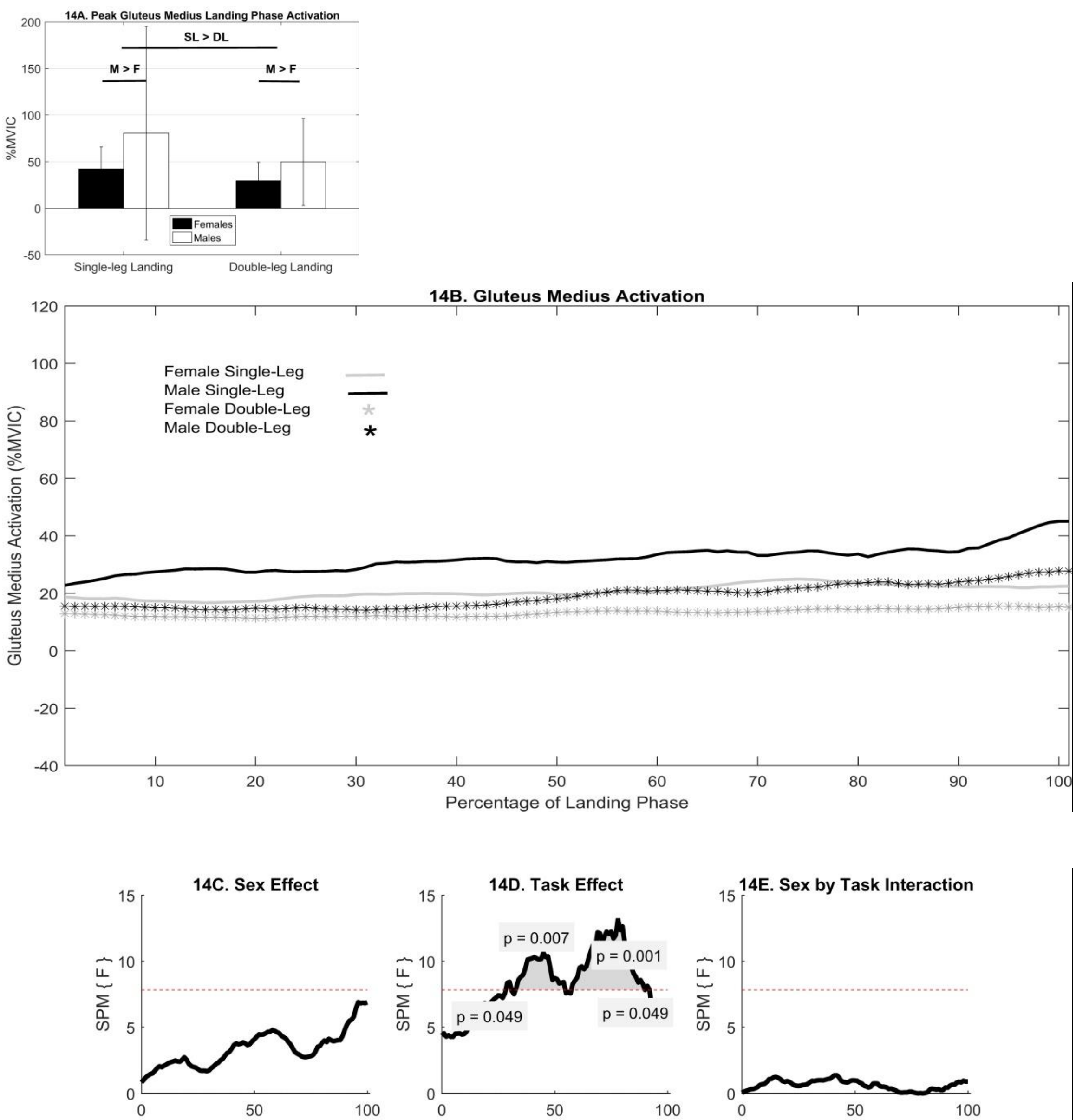
Figure 4.13. Gluteus Maximus EMG Amplitude (%MVIC): Univariate Results (13a) and SPM Descriptive and Inferential Results (13b-e).



Gluteus Medius Activation. Univariate analysis revealed main effects for sex (p=.01) and task (p=.01) for gluteus medius muscle activation. Regardless of task, females utilized 29.4% less of their gluteus medius than did males (d=.26) (Figure 4.14a). Regardless of sex, gluteus medius activation was 21.8% greater during $LAND_{SL}$ (d=.33) than $LAND_{DL}$ (Figure 4.14a).

SPM analysis revealed main effects for task ($F_{crit(1,88)}$=7.84), but not for sex. The greater gluteus medius muscle activation during $LAND_{SL}$ compared to $LAND_{DL}$ occurred from 29-30% (p=.05), 33-54% (p=.01), 58-89% (p=.001), and at 91% (p=.05) of the landing phase (Figure 4.14d).

Figure 4.14. Gluteus Medius EMG Amplitude (%MVIC): Univariate Results (14a) and SPM Descriptive and Inferential Results (14b-e).



Discussion

We hypothesized that, compared to males, females would exhibit greater functional valgus collapse, operationally defined as the angles and joint moments associated with greater knee abduction and internal rotation and hip adduction and internal rotation, and that this effect would be more pronounced in the $LAND_{SL.}$ Our results partially supported this, with results generally revealing that sex differences in frontal plane knee movement were task dependent. During the $LAND_{SL}$, females displayed greater knee abduction than males. This was particularly true during the early stages of the landing phase, as evidenced by SPM results. At the hip, females maintained greater hip adduction throughout both tasks, while also utilizing a lower percentage of their gluteus medius than did males. The latter was contrary to our hypothesis that females would use a larger proportion of their available gluteal torque generation capabilities to complete the $LAND_{SL}$ and $LAND_{DL}$ given their lower normalized torque producing capabilities compared to males. However this unexpected result was limited to the gluteus medius, and was not confirmed using SPM analysis. Moreover, when comparing results obtained from the GLM vs SPM, we gained clarification and more detailed information of where in the landing phase these sex, task, and sex by task differences were occurring. The following sections will discuss each of these points further.

Sex Differences that were Task Dependent. As demonstrated by the sex by task interactions we observed at the knee in all three planes of motion and at the hip in the transverse plane, the demands of the task can moderate the presence or magnitude of observed sex differences. This is a salient consideration for clinicians in choosing appropriate screening tests, and in the development of sex and activity specific injury prevention programs. In the current study, this was particularly apparent in frontal plane knee movement. Knee abduction is a primary component of functional valgus collapse (Ireland, 1999) and is often referenced as an indicator of ACL injury risk, particularly in females (Barry P Boden et al., 2009; Boden BP, Dean GS, Faegin JA, 2000; Timothy E Hewett et al., 2005; OKane et al., 2016). In the current study, visual inspection of time series curves suggests that male frontal plane knee excursions are minimal when compared with female motion throughout both tasks (Figure 4.2f). However these sex differences were more apparent in the $LAND_{SL}$, where females' initial knee abduction was more affected than males'; females were 2.3° more abducted in the $LAND_{SL}$ than in the $LAND_{DL}$, whereas males were only 0.8° more abducted in the $LAND_{SL}$ (GLM analysis). This interaction may be more important when viewed in light of the SPM task effect, which revealed greater knee abduction during the first third of the $LAND_{SL}$ than in the $LAND_{DL}$ (Figure 4.2f & 4.2h); this is the time during which ACL injuries are thought to occur (T Krosshaug et al., 2007). Hence, the amplified sex difference in initial valgus angle during the $LAND_{SL}$ may put females in a more precarious position during the early phases of single-leg landing tasks. Knee flexion angle during this time is also a pertinent consideration, as smaller knee flexion angles encourage knee valgus (Berns et al., 1992; Fukuda et al., 2003). Thus, it is possible that greater knee abduction during the first third of the landing phase when combined with smaller knee flexion angles in females could create a particularly injurious scenario.

While females' initial knee adduction angles were more negatively influenced by the $LAND_{SL}$, their knee abduction excursions were more pronounced in the $LAND_{DL}$; females moved through 0.7 more degrees than males in $LAND_{SL}$, but 3.3° greater motion than males in the $LAND_{DL}$. This resulted in greater absolute peak knee abduction during the $LAND_{DL}$ than in the $LAND_{SL}$ ($\Delta = 3.6°$). While one may be tempted to interpret these findings that $LAND_{DL}$ is more detrimental for females, it is also important to consider when this peak knee abduction is occurring during the landing phase. In the current study, the average time for completion of the functional task was 207.50 ± 38.17ms ($LAND_{SL}$: 205.6 ± 34.5ms; $LAND_{DL}$: 209.4 ± 41.8ms). During the $LAND_{DL}$, peak knee abduction occurred at approximately 75-80% of the landing phase, or 155-166ms after initial ground contact. During the $LAND_{SL}$, peak knee abduction occurred at approximately 15-20% of the landing phase, or 31-42ms after initial ground contact (Figure 4.2f). It is accepted that the 30-60ms time frame after initial ground contact is most critical to ACL injury occurrence (Carlson et al., 2016; T Krosshaug et al., 2007; Tron Krosshaug et al., 2007). Therefore, even though peak knee abduction is greater during the $LAND_{DL}$, the lesser and earlier peak during the $LAND_{SL}$ may be more paramount. Our SPM analysis supports this. From initial ground contact until 75ms (0-36%) after contact, $LAND_{SL}$ knee abduction is approximately 2-3° greater than knee abduction during the $LAND_{DL}$ (Figure 4.2h). Coupled with previous research documenting that 72% of ACL injuries occur during a single-leg stance (Barry P Boden et al., 2009), this information suggests that elevated knee abduction during the first 75ms of the landing phase may possibly contribute to the injurious nature of single-leg activities more so than knee abduction during the latter stages of landing. Therefore, despite the $LAND_{DL}$ displaying greater absolute peak knee abduction angles, the peak knee abduction observed earlier in the landing phase of $LAND_{SL}$ may be more problematic, especially in a female population already displaying elevated knee abduction at initial ground contact of the single-leg landing. These findings suggest that researchers and clinicians aiming to assess the amount of valgus collapse in females should carefully consider the activity during which this pattern is being observed, and how the task may impact observed sex differences and their implications relative to injury risk potential. It has been previously noted that frontal plane kinematics are critical to ACL injury. The vertical drop jump task that is commonly used to assess frontal plane knee kinematics may not be challenging enough, as it has displayed poor sensitivity and specificity when discriminating between ACL-injured and healthy individuals (T. Krosshaug et al., 2016).

Therefore, a single-leg task that is more challenging for the musculature could be a more appropriate screening task, though more research is needed to confirm the efficacy of a single-leg landing task for ACL injury screening. Moreover, interventionists should be mindful of task differences when tailoring injury prevention programs for females engaging in activities with particular single-leg or double-leg demands.

Sex Difference Main Effects. In addition to interactions for frontal plane knee movement, we also identified main effects for sex in peak knee abduction and frontal plane hip motion that can contribute to dynamic knee valgus. An interaction was not identified, indicating that task did not moderate sex differences in peak knee abduction and frontal plane hip motion. In the current study, females' peak knee abduction angles were 4.1° larger than males'. This agrees with a 2016 systematic review which concluded that females display greater peak knee abduction angles than males across various weight-bearing tasks (Cronström, Creaby, Nae, & Ageberg, 2016a). The timing of our observed peak knee abduction is also an important consideration. Not only did females display 4.1° more knee abduction than males, but SPM revealed this difference to occur during the 37-46% section of the landing phase (Figure 4.2g). Previous research demonstrated that during jumping activities, such as those performed in basketball, peak vertical ground reaction force (GRF) occurred approximately 30-40 milliseconds (ms) after initial ground contact, followed by peak anterior-posterior GRF at 60-100ms post-initial ground contact (T Krosshaug et al., 2007). Simultaneous with peak GRFs, knee abduction dramatically increased and remained elevated until approximately 130ms after initial ground contact (T Krosshaug et al., 2007). The 37-46% section in which we observed greater knee abduction in females would on average correspond to the time frame of 76-95ms post-initial ground contact. While this time frame is beyond the critical 30-50 ms window, our observed increase in female knee abduction could be reflective of a retrospective spike in vertical GRF. Though our data didn't inferentially support this, visual inspection of the SPM data (Figure 4.8c) indicates that females did maintain greater knee abduction moments throughout both landing tasks than did males. Thus, in addition to the absolute differences in peak knee abduction angle identified by GLM, the timing and duration of knee abduction established by SPM may be an important factor to consider when assessing the presence of *injurious* functional valgus collapse.

Previous research has indicated that frontal plane hip motion couples with frontal plane knee motion during single-leg cutting maneuvers, accounting for as much as 25% of the variance in knee abduction angles (Imwalle, Myer, Ford, & Hewett, 2009). In the current data, a post-hoc bivariate correlation revealed that peak hip adduction accounted for 37% of the variance in peak knee abduction during the $LAND_{SL}$ ($p < .001$). Also, females displayed 2.1° greater peak hip adduction and 1.9° greater hip internal rotation than males. It's possible that these differences contributed to the 4.1° difference in peak knee abduction (Hollman, Galardi, Lin, Voth, & Whitmarsh, 2014; Imwalle, Myer, Ford, & Hewett, 2009). Although observed sex differences in frontal plane hip movement were minimal, it still informs the biomechanical sex disparity, due to the hip-knee coupling in the frontal plane and the excessive frontal plane hip motion during the $LAND_{SL}$. Visual inspection of the time series curves (Figure 4.5f) reveals that during the $LAND_{SL}$, the hip is freer to adduct; whereas during the $LAND_{DL}$ there is minimal frontal plane hip movement. Our data empirically support this idea. Initial hip abduction (Figure 4.5a), peak hip adduction (Figure 4.5b), peak hip abduction (Figure 4.5c), and hip adduction excursion (Figure 4.5d) were all greater in the $LAND_{SL}$. SPM confirmed the more extreme nature of frontal plane hip motion during the $LAND_{SL}$. During the first third of the $LAND_{SL}$ (0-33%), participants' hips were more abducted. This is reversed during the latter half of the landing phase (49-100%), when participants displayed greater hip adduction in the $LAND_{SL}$ (Figure 4.5h) than in the $LAND_{DL}$. With only one limb contacting the ground, frontal plane hip movement was encouraged during a $LAND_{SL}$, while the apparent splinting of the joint during the $LAND_{DL}$ was likely a result of using dual frontal plane supports. Though intuitive, this distinction is nonetheless important. In order to remain upright during a single-limb stance, the body's center of mass must shift laterally, which then initiates a reactive hip adductor moment (Barry P Boden et al., 2009; Timothy E Hewett & Myer, 2011). Because of the hip's frontal plane influence over the knee, the lack of hip control in a single-leg stance could be more problematic for females, seen in light of the previously discussed sex differences in knee abduction motion. Moreover, the lone sex difference in frontal plane hip movement was hip abduction excursion. Visual inspection of the descriptive SPM graph (Figure 4.5f) suggests this difference to stem from the first 20% of the landing phase, which corresponds to the greater knee abduction females also displayed during this time frame, thus further reinforcing a hip-knee coupling concept.

As hip adduction during the $LAND_{SL}$ began to increase at approximately 30% (Figure 4.5f), gluteus medius activation also increased in a similar fashion (Figure 4.14b). This suggests that a single-leg landing task taxes the gluteus medius to a higher degree, perhaps suggesting a lack of hip control and exaggerated frontal plane hip motion in those with inferior strength and activation. This point is illustrated further when considering that, with the exception of hip adduction, participants landed more stiffly during the $LAND_{SL}$. This is particularly true with knee (Figure 4.1) and hip flexion (Figure 4.4), but is also apparent in knee ab/adduction (Figure 4.2) and hip rotation (Figure 4.6). This suggests that frontal plane hip movement accounted for a large proportion of total lower extremity motion during the $LAND_{SL}$. The generous hip adduction excursion, combined with small excursions in the sagittal and transverse planes, may have further increased demands on the gluteus medius to control hip adduction. Given such a movement strategy, an efficient gluteus medius may be imperative to controlling frontal plane hip movement, especially during single-leg activities. As expected, males had stronger gluteal muscles than females when normalized to body mass. This is consistent with prior work (Jacobs et al., 2007; Willson et al., 2006). Additionally, males used a greater proportion of their gluteus medius than females during both functional tasks, as borne out by GLM analysis. This was unexpected, as we postulated that weaker females would necessarily recruit a greater proportion of available strength when completing a similar task as stronger males. However, males also maintained approximately 2-3° more hip abduction throughout both functional tasks, so it is possible that the increased neural drive to the gluteus medius enabled males to adopt a safer movement strategy. This is in contrast to the greater hip adduction and decreased gluteus medius activation observed in females, suggesting that a single-leg landing task may pose more of a neuromechanical challenge for females. As such, further research is warranted to determine if the gluteus medius represents a potential avenue whereby frontal plane hip motion, thus frontal plane knee motion, can be controlled.

These findings may suggest multiple avenues whereby intervention may be possible and highlight directions for future research. Future work should determine the efficacy of preparatory action prior to initial ground contact, and whether it is possible to partially mitigate one's propensity for high risk biomechanics by limiting knee abduction at ground contact. This may require interventions aimed at the hip specifically, to ensure the hip does not fall into adduction and internal rotation, which would increase the potential for greater initial knee abduction (Hollman et al., 2014) and greater valgus collapse.

GLM v. SPM. A summary of results yielded by each analysis is presented in Table 4.6. GLM analysis and SPM analysis are complementary to each other, each having their own purpose and providing unique information. SPM is a node by node analysis, and is not intended to infer slope relationships. For this reason, SPM is not effective in identifying effects related to joint excursions. Therefore, if joint excursions are vital to the research question, GLM analysis may be a more suitable option. However, if one is interested in knowing when in the landing phase sex and task differences are occurring relative to injury risk potential, SPM may be the more appropriate method. Take for instance the previously discussed between-task difference in knee abduction patterns. If only the GLM results had been considered, not taking into account the timing difference between $LAND_{SL}$ and $LAND_{DL}$ peak knee abduction, one would think that double leg landings are more dangerous for females. However, when examining the timing of these differences in $LAND_{SL}$ and $LAND_{DL}$, the greater valgus in females occurred much earlier in $LAND_{SL}$, which corresponds to what is known about peak ground reaction forces and peak ACL strain (Kiapour et al., 2014). In the $LAND_{DL}$, peak knee abduction occurs much later, and also with greater knee flexion (considered a safer position) (Fukuda et al., 2003). Therefore, even though knee abduction appears greater in $LAND_{DL}$, it is not as telling as the knee abduction occurring in $LAND_{SL}$. This is a critical component to the development of effective screening and intervention strategies. Given these substantial differences in movement patterns between tasks, and that the majority of ACL injuries occur in a single-limb stance (Barry P Boden et al., 2009), prevention programs should emphasize single-leg movement quality with a focus on the first third of the landing phase. Additionally, GLM and SPM also differed on gluteus medius activation. GLM identified a sex difference for peak gluteus medius amplitude, whereas SPM identified no differences throughout the landing phase. Although the SPM results clearly show that males have greater muscle activation during $LAND_{SL}$ (Figure 4.14b), the null results are likely due to the high variability of male, particularly during the single-leg task. In this case, using both analyses is useful to verify the robustness of the results. If one analysis is unduly driven by skewed variability, clinicians have the option to more closely inspect the findings and use their best judgment in reaching an informed clinical decision.

Table 4.6. A Summary of Significant (p<.05) Results Yielded from the General Linear Model (GLM) and Statistical Parametric Mapping (SPM) Analyses.

	GLM			SPM		
Variable	Sex	Task	Int.	Sex	Task	Int.
Knee Flexion Kinematics	--	Exc	IC Pk	--	0-100%	0-1% 2-10% 11-100%
Knee Frontal Plane Kinematics	Pk (Ab)	Pk (Ab)	IC Exc (Ab)	37-46%	0-36% 54-100%	--
Knee Transverse Plane Kinematics	--	IC	Exc (ER)	--	0-5%	--
Hip Flexion Kinematics	IC Pk	Pk Exc	--	0-14% 45-100%	28-39% 44-100%	--
Hip Frontal Plane Kinematics	Exc (Ab)	IC Pk (Ad) Pk (Ab) Exc (Ad)	--	--	0-33% 49-100%	--
Hip Transverse Plane Kinematics	IC Exc (ER)	Pk (ER) Exc (ER)	Exc (IR)	0-2%	52-76%	--
Knee Flexion Kinetics	--	--	--	0-1%	0-1% 3-52%	--
Knee Frontal Plane Kinetics	--	Pk (Ad)	--	--	0-100%	--
Knee Transverse Plane Kinetics	--	Pk (IR)	--	--	22-100%	--
Hip Flexion Kinetics	--	--	--	0-1% 73-85%	0-27%	48-56%
Hip Frontal Plane Kinetics	--	Pk (Ad)	--	--	0-100%	--
Hip Transverse Plane Kinetics	--	Pk (ER)	--	--	4% 52-100%	--
Gluteus Maximus	--	--	--	--	--	--

Activation							
Gluteus Medius Activation		Pk	Pk	--	--	29-30% 33-54% 58-89% 91%	--

Percentages are referenced to the landing phase; IC=joint angle at initial contact; Pk=peak joint angle; Exc=joint excursion; int.=interaction; Ab=abduction; Ad=adduction; ER=external rotation; IR=internal rotation

Limitations. The authors acknowledge the existence of limitations in the study design and analyses used. It is known that sagittal plane hip and knee position influence one's potential to display functional valgus collapse (Delp, Hess, Hungerford, & Jones, 1999; Fukuda et al., 2003; van Arkel et al., 2015); knees flexed less than 30° are more prone to valgus forces (Fukuda et al., 2003), and greater degrees of hip flexion encourage the hip to externally rotate (van Arkel et al., 2015). Using a MANCOVA to control for sagittal plane hip and knee position would have been more ideal. However, not only would this have been cumbersome statistically and for interpretation, but it would not have allowed for a true comparison between analyses, given that SPM does not support control variables. Instead, we have presented the full sagittal plane MANOVA and SPM results, thus allowing the reader to make well-informed inferences as to the robustness of our interpretations. Future work could address this limitation by quantifying the influence of sagittal plane hip and knee position on functional valgus collapse via their inclusion as control variables in either a group comparison or correlative analysis.

In order to accurately compare time series curves between sexes and tasks, it was necessary to standardize the landing phase lengths to 100%. The time it took for task completion was variable, and we acknowledge that registering the time series curves to 100% may have masked a portion of inter-subject variability, as well as the absolute timing of when differences occur, which may be relevant to injury risk potential. However, as there were no significant differences in landing phase lengths between sexes or tasks, normalization to 100% likely did not have an appreciable effect on our significant findings, as it would have added random and not systematic errors. Future research could avoid this by analyzing a pre-determined number of milliseconds, instead of the entire landing phase.

Lastly, because of the relatively homogenous sample, our results may not be generalizable to populations other than young healthy adults, though our sample was taken from a population in which ACL injuries commonly occur.

Conclusion

In conclusion, there were substantial differences in frontal plane hip and knee motion between sexes, particularly in the $LAND_{SL}$, where females exhibited greater knee abduction in the first third of the task, along with greater hip adduction and decreased gluteus medius activation. Coupled with the finding that females exhibited greater knee abduction than males, particularly from 37-46% of the landing phase, the increased knee abduction angle at initial contact may put females in particularly compromising situations during the early stages of single-leg landings. This finding may be in part due to males using a greater proportion of their gluteus medius than did females, which also corresponded with greater hip abduction in males in both tasks. To confirm this relationship, future studies can use an SPM correlative analysis (e.g. regression, canonical correlation) to determine the exact time frames at which the gluteus medius most strongly influences hip adduction. This would determine the potential for the gluteus medius to be an effective intervention target for controlling excessive hip adduction. Frontal plane knee and hip motion was substantially different between tasks, where gluteus medius activation was substantially greater and hip adduction was more extreme throughout the $LAND_{SL}$, and abduction was substantially greater during the first third (0-36%) of the $LAND_{SL}$ compared to $LAND_{DL}$. Thus, further research is needed to examine the relative timing of these events, and the extent to which gluteus medius activation may be trained to effectively control hip and knee frontal plan motions. Depending on the research question, Statistical Parametric Mapping is useful as a stand-alone analysis or when used in conjunction with more conventional statistics to provide a more complete description of biomechanical patterns, particularly relative to timing in the landing phase, and the timing of movement between planes and joints. Researchers should conscientiously choose the analysis (or combination of analyses) that best answers their specific research questions. Lastly, when assessing the amount of functional valgus collapse in females, both researchers and clinicians should be cognizant of how the chosen activity type and the statistical approach taken can significantly alter observed biomechanical effects.

CHAPTER V

MANUSCRIPT II. THE EFFECTS OF GLUTEAL STRENGTH AND ACTIVATION ON THE RELATIONSHIP BETWEEN FEMORAL ALIGNMENT AND FUNCTIONAL VALGUS COLLAPSE

Abstract

Context. An anatomical bias toward femoral internal rotation is a potential precursor to functional valgus collapse, a potential risk factor for ACL injury. Gluteus maximus and medius, which stabilize the hip during stance and control hip adduction and internal rotation, may play a critical role in mitigating the effects of sub-optimal femoral alignment.

Objective. Determine the extent to which gluteal muscle strength and activation influence the associations between femoral anteversion and passive internal and external rotation hip ROM (ROM_{IR} and ROM_{ER}) with functional valgus collapse during a single-leg forward landing task. We hypothesized that greater femoral anteversion and greater ROM_{IR} and lesser ROM_{ER} would predict increased joint angles and external moments associated with knee abduction, knee internal rotation, hip adduction, and hip internal rotation. We also hypothesized that the strength of these relationships would decrease, but the overall prediction (R^2) model would be strengthened once accounting for the influence of gluteus maximus and gluteus medius strength and activation.

Design. Cross-sectional.

Setting. Research laboratory.

Patients or Other Participants. Forty-five females (20.1±1.7 yrs, 165.2±7.6 cm, 68.6±13.1 kg) and forty-five males (20.7±2.0 yrs, 177.7±8.5 cm, 82.8±16.3 kg) aged 18-25, with no history of lower extremity surgery and no injury history in the previous six months.

Intervention(s). Femoral anteversion and passive hip ROM were measured prone with the knee at 90°. Maximal voluntary isometric contractions were obtained for hip extension and abduction. Three-dimensional biomechanics and surface electromyography were obtained during performance of single-leg forward landing tasks.

Main Outcome Measures. Forward-stepwise multiple linear regressions were used to determine the influence of femoral anteversion, ROM_{IR} and ROM_{ER} on initial and peak joint angles, joint excursions, and peak external joint moments (first step) and the mediating effects of gluteus maximus and gluteus medius strength (MVIC) and activation (EMG amplitude as a percentage of MVIC) (second step).

Results. In females, femoral alignment predicted knee rotation angles (R^2 range = .11-.28, p range = .02-.31). Increased gluteus maximus activation (part r range=.32-.37) strengthened the part correlations between femoral alignment and knee rotation (femoral anteversion Δr=.07, ROM_{IR} Δr range =.09-.11) as well as the overall prediction of greater initial (R^2=.21, p=.09) and peak (R^2=.42, p=.001) knee internal rotation. Though no anatomical variables predicted frontal plane hip motion, less hip abduction strength (part r = -.32) predicted greater peak hip adduction (R^2=.28, p=.003), while greater hip extension strength (part r = -.29) predicted greater peak hip abduction (R^2=.21, p=.02).

In males, femoral alignment predicted knee rotation angles (R^2 range = .13 - .33, *p* range = .01 - .18) and knee abduction moment (R^2=.18, *p*= .08). The addition of less hip abduction strength (part *r* range= -.31- -.41) strengthened the overall prediction of greater initial (R^2=.25, *p*=.02) and peak (R^2=.39, *p*=.001) knee internal rotation, lesser peak knee external rotation (R^2=.26, *p*=.03), and lesser peak knee abduction moment (R^2=.28, *p*= .02), while weakening the part correlations between anatomy and knee rotation (femoral anteversion Δr range=.02-.06, ROM_{IR} Δr= -.15) and strengthening the relationship between anatomy and frontal plane knee moment (ROM_{IR} Δr= -.10, ROM_{ER} Δr= -.06). Greater hip extension strength (part *r*= .29) and less gluteus maximus activation (part *r*= -.25) strengthened the overall prediction of less hip external rotation moment (R^2=.47, *p*= .001), while the part correlations between anatomy and hip rotation moment remained constant (Δr range= -.02-.01).

Conclusions. In females, a bias toward internal hip rotation (greater ROM_{IR} and lesser ROM_{ER}) was predictive of variables associated with greater functional valgus collapse. In males, greater femoral anteversion and more flexible hips (greater ROM_{IR} and greater ROM_{ER}) predicted greater functional valgus collapse. Gluteal function was useful for explaining additional variance in hip and knee biomechanics, but often did not alter the relationship with anatomy. Further research is needed to determine the extent to which gluteal neuromuscular interventions affect biomechanical outcomes in males and females, and the extent to which passive hip ROM is a modifiable risk factor.

Key Words. ACL, hip ROM, femoral anteversion, gluteal activation, valgus

Introduction

Over the previous two decades, copious research has worked to identify risk factors predisposing one to functional valgus collapse, a high-risk biomechanical movement strategy thought to contribute to anterior cruciate ligament (ACL) rupture (Timothy E Hewett et al., 2005; OKane et al., 2016). Functional valgus collapse is more often observed in females (B P Boden, Torg, Knowles, & Hewett, 2009; T E Hewett et al., 2009; T Krosshaug et al., 2007), and describes the combined lower extremity motions of hip adduction and internal rotation, and knee abduction and internal rotation (Ireland, 1999; Meyer & Haut, 2008). To explain the sex disparity regarding movement patterns, researchers have sought to identify anatomical characteristics which may predispose females to display greater valgus collapse. Because a valgus collapse strategy consists of coupled movements between the hip and knee (Hollman et al., 2014; Imwalle, Myer, Ford, & Hewett, 2009), anatomical characteristics affecting the hip may influence one's propensity for functional valgus collapse.

The anatomical characteristics of femoral anteversion and passive hip internal rotation range of motion (ROM) are greater in females (A.-D. Nguyen & Shultz, 2007; Hogg et al, in press) and are separately thought to contribute to greater movements toward functional valgus collapse by predisposing the hip toward greater adduction and internal rotation (Howard, Fazio, Carl, et al., 2011; A.-D. Nguyen et al., 2015). While femoral anteversion is characterized by a medially-torsioned femoral neck and is developmental (Fabry, 1973; Kozic et al., 1997), passive hip ROM is a pliable soft tissue restraint, which may be influenced by other factors such as general joint laxity (Fan et al., 2014a), which has been shown to remain constant throughout maturation (S. J. Shultz et al., 2008). Though these two characteristics may be somewhat distinct, together they have the potential to influence functional valgus collapse. While these factors may be difficult to modify, it is still important to understand their influence on lower-extremity biomechanics, so that we can better identify those at risk and design effective prevention strategies.

To that end, the gluteus maximus and gluteus medius muscles may represent effective modifiable intervention targets to mitigate the negative effects of sub-optimal femoral anteversion or passive hip ROM. As primary external rotators and abductors of the hip, these muscles work eccentrically to control dynamic hip internal rotation and adduction, two primary components of functional valgus collapse. This is particularly true in a single-leg stance, when the gluteals are more challenged to maintain a level pelvis. Moreover, greater degrees of femoral anteversion or passive hip internal rotation ROM result in lengthened moment arms for the gluteus maximus and medius (Free & Delp, 1996; J Nyland et al., 2004; Radin, 1979), thus potentially affecting their torque generation capabilities. To this point, research suggests that greater femoral anteversion and internal rotation ROM may each be associated with decreased gluteal efficiency (Howard, Fazio, Mattacola, et al., 2011; Kaneko & Sakuraba, 2013a; J Nyland et al., 2004; Sigward et al., 2008), as evidenced by decreased torque generation (MVIC) coupled with increased EMG amplitude (Homan et al., 2013a; A.-D. Nguyen et al., 2011). This neuromuscular profile is suggestive of an individual with weaker and less effective gluteals, who must use a greater proportion of their strength to complete a given task. In so doing, the muscle could more readily fatigue, and be unable to affect safe movement strategies. While addressing gluteal strength and activation may represent a viable intervention target in individuals displaying high femoral anteversion or passive internal rotation ROM, the extent to which strength and activation of the gluteals can counteract problematic femoral alignment, or their function compromised by alignment, has not yet been elucidated.

Therefore, the objective of this study was to determine the extent to which gluteal muscle strength and activation mediate associations between femoral anteversion and passive internal and external rotation hip ROM with functional valgus collapse during a single-leg forward landing task in females and males. We hypothesized that increased femoral anteversion and internal rotation ROM and decreased external rotation ROM would predict greater hip and knee movement toward functional valgus collapse during a single-leg forward landing task. We further hypothesized that the addition of increased gluteal activation with decreased hip strength would strengthen the overall prediction of the biomechanical variable, while weakening the relationship between femoral alignment and functional valgus collapse.

Methods

Participants. Forty-five female participants (20.1±1.7 yrs, 165.2±7.6 cm, 68.6±13.1 kgs) and forty-five male participants (20.7±2.0 yrs, 177.7±8.5 cm, 82.8±16.3 kgs) were recruited for a single session. This sample size was based on an *a priori* power analyses of pilot data in a female cohort, and determined to be adequate to detect moderate effect sizes with a power of .80. To ensure a homogenous sample, specific inclusion criteria were 1) adults between the ages of 18 and 25 and 2) a score of two or more (at least "one time in a week") on categories 2-4 ("cutting, "decelerating", and "pivoting") of the Marx activity rating scale (see Appendix). Specific exclusion criteria were 1) any history of knee surgery, 2) any history of ligamentous or meniscal knee injury, 3) any history of lower extremity injury within the previous 6 months, 4) history or diagnosis of a vestibular condition affecting balance, and 5) history or diagnosis of any cardiovascular condition precluding exercise. Each participant provided written informed consent as approved by the university's Institutional Review Board. After obtaining written informed consent, each participant was asked to complete the following intake questionnaires: Physical Activity and Health History Questionnaire, Knee Outcome Survey (both the Activities of Daily Living Scale and the Sports Activities Scale), and the Marx Activity Rating Scale (Appendix A).

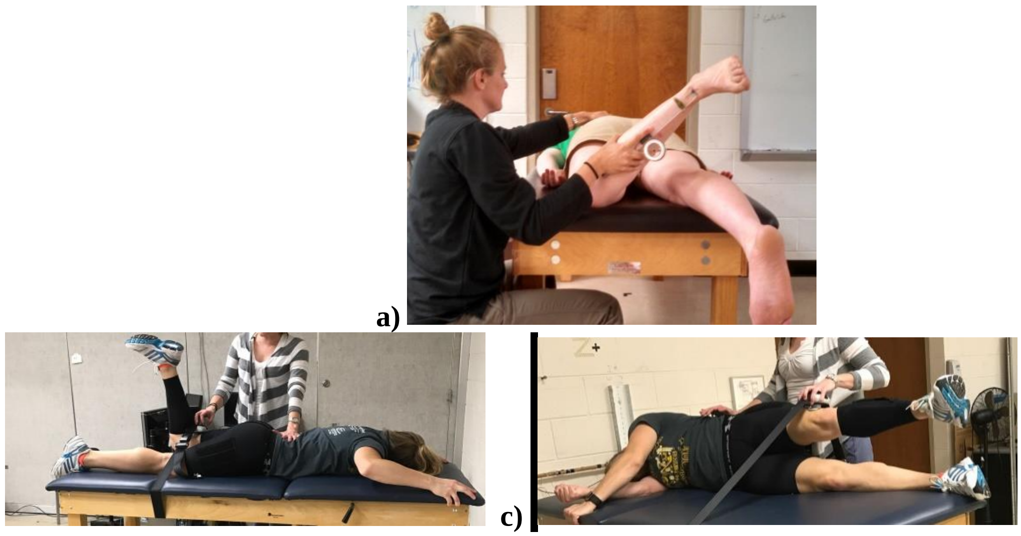
Anatomical Measures. Both femoral anteversion and hip ROM were measured prone with a standard inclinometer with the hip in neutral and the knee flexed to 90° (Magee, 1997). To measure femoral anteversion, the examiner rotated the lower leg while palpating the greater trochanter. When the greater trochanter was at its most lateral point, the transverse plane angle formed by the tibial shaft and true vertical was measured as femoral anteversion. To measure hip ROM, the examiner passively internally and externally rotated the lower leg while palpating the sacrum. At the point of initial sacral movement, the transverse angle formed by the tibial shaft and true vertical was measured as ROM_{IR} and ROM_{ER}, respectively (Figure 5.1a). Three trials were taken for each measure and averaged for analysis. The same examiner took all measurements, and had previously established good to excellent inter-day reliability [$ICC_{2,3}$(SEM); femoral anteversion: .92(1.2°); internal rotation ROM: .97(1.6°); external rotation ROM: .85(3.3°)] (Hogg, Schmitz, Nguyen, & Shultz, in press).

Surface Electromyography Instrumentation. Each participant was outfitted with surface electromyography (EMG) double differential electrodes (Trigno Wireless Sensors, Delsys, Boston, MA) to acquire signals from the gluteus medius and gluteus maximus during maximal strength testing and performance of the $LAND_{SL}$. Prior to sensor placement, the skin was cleaned with an alcohol swab. The gluteus medius electrode was placed one-third the distance from the iliac crest to the greater trochanter (Rainoldi et al., 2004). The electrode on the gluteus maximus was placed halfway between the second sacral vertebrae to the greater trochanter (Rainoldi et al., 2004). All electrodes were positioned parallel to muscle fiber orientation and secured with tape or prewrap. Proper positioning was verified with manual muscle testing and visual inspection of the EMG signal using EMGWorks (Delsys, Boston, MA). EMG data were sampled at 1000Hz and were manually synced with kinematic and kinetic data during post-collection processing.

Maximal Voluntary Isometric Contractions (MVICs). Prior to obtaining MVICs, each participant completed a five minute warm up on a stationary bike at a self-selected pace. Following warm-up, MVICs of the gluteus maximus (hip extension) and gluteus medius (hip abduction) were obtained and used as representations of maximum torque generation values, as well as for normalization of EMG amplitude. For all MVIC measures, a strap was used to secure the dynamometer in place and to provide resistance for the participant. For MVIC measurement of the hip extensors, the participant was positioned prone with the knee bent to 90 degrees. With a handheld dynamometer (Lafayette Instruments, Boston, MA) placed over the posterior distal thigh two inches proximal to the joint line, the participant was asked to maximally contract into hip extension (Bohannon, 1986; Starkey & Ryan, 2003). Hip abduction MVICs were measured side-lying on the right side, with the left leg up. The left leg was placed in 10-15 degrees of hip extension and slightly externally rotated, thus isolating the gluteus medius. Maximal hip abduction was resisted by placing the lower edge of the dynamometer two inches proximal to the lateral knee joint line (Krause et al., 2007). Prior to collecting each MVIC, participants were familiarized to the measure and allowed up to three submaximal practice trials. Each condition consisted of three 5-second trials, with 30 seconds rest between trials. To prevent an artificial spike in dynamometer output during collection, each participant was instructed to slowly increase force output, reaching maximum force production during the third second of the five second trial.

Prior to the study the investigator established reliability for strap-assisted handheld dynamometry [$ICC_{2,2}$(SEM); hip extension: .76(3.4kg); hip abduction: .96(1.6kg)].

Figure 5.1. Measurement Position of Anatomical Variables (a), Hip Extension MVIC (b), and Hip Abduction MVIC (c).



Biomechanical Instrumentation. Prior to digitization for motion capture, participants were outfitted with standardized shoes (Adidas Uraha 2, Adidas AG, Herzogenaurach, Bavaria) to eliminate potential shoe-surface interactions. Participants were adorned with six marker clusters, placed at each of the following locations: lateral aspect of the left foot, lateral aspect of the left lower leg (mid-shaft), medial and lateral proximal tibial flares, the lateral left thigh (mid-shaft), the L5-S1 junction, and the postero-superior thorax (C7-T1 spinous processes). Participants were then digitized using the MotionMonitor software (Innovative Sports Training, Chicago, IL). Joint centers for the knee and ankle were defined as the midway point between the medial and lateral femoral condyles and medial and lateral malleoli, respectively. The hip joint center was determined using the Bell method (A. L. Bell & Pedersen, 1989). A segment-based coordinate system was used to define each body segment. The X-axis was defined as the anterior-posterior axis (adduction/abduction), the Y-axis was the distal-proximal axial axis (internal/external rotation), and the Z-axis was defined as the medial-lateral axis (flexion/extension). Motions for each joint were calculated using Euler's equations (Z Y' X") (Kadaba et al., 1989). Kinematic data were obtained with an 8-camera optical LED system (Impulse, Phase Space; San Leandro, CA) at a sampling rate of 240 Hz. Kinetic data were obtained using an embedded Bertec forceplate (Type 4060-130; Bertec Corporation, Columbus OH, USA) and were sampled at a rate of 1000 Hz. Kinematic and kinetic instrumentation were interfaced with MotionMonitor software and were manually synced by a pulse trigger during each trial.

Procedure for Single-Leg Forward Landing. Participants were familiarized to the task prior to data collection and were allowed up to three practice trials. Tape was placed on the ground at a distance equal to 40% of each participant's height away from the front edge of the forceplate. A foam barrier equal to 15% of the participant's height (Jacobs et al., 2007) was placed halfway between the tape and the edge of the forceplate. Instructions to participants were standardized. Participants were instructed to stand behind the tape and jump from 2 legs over the foam barrier, landing on their left foot (Figure 5.2). Trials were discarded if the participant double-hopped upon landing, hit the barrier, didn't clear the barrier with both feet, didn't land with the entire left foot on the forceplate, or used their contralateral limb for additional support. Five clean trials were collected and used for analysis.

Figure 5.2. Terminal Position for the Single-Leg Forward Landing Task.

Data Handling and Processing. Motion capture began two seconds before initial ground contact, defined as the point at which the vertical ground reaction force exceeded 10N, and continued for three seconds after initial ground contact (total five seconds). All flexions, adductions, and internal rotations were defined as positive, and all extensions, abductions, and external rotations were defined as negative angles. To determine the appropriate filter for kinematic and kinetic data, a residual analysis was conducted on the ground reaction force (GRF) in a subset of the trials. Because the signal to noise ratio is dependent on the physical motion capture system and the velocities being captured and should be stable across participants, using a subset of the data is more than adequate to determine the appropriate filtering frequency. (E. Kristianslund et al., 2012; Eirik Kristianslund et al., 2012; Winter, 1990). To conduct the residual analysis, raw trials from four randomly selected participants were used. Each trial was filtered in MATLAB (MathWorks, Inc., Natwick, MA) under a series of low-pass filters, yielding multiple "versions" of each GRF time series. A sum of squares was computed for each non-raw version at each time point. This can be represented as $(\text{raw}_X\text{-filtered}_{yx})^2$, where x is the frame of data and y is the filtering frequency. Once the sums of squares were computed for each frequency, residuals were obtained. For each filtering frequency, the residual is defined as the square root of the mean sum of squares across all time points (Winter, 1990). Finally, to determine the proper filtering frequency, separate plots were generated for each GRF residual. Visual inspection of the plots indicated that the optimum ratio of signal distortion to noise was approximately at 10 Hz. Therefore, all kinetic and kinematic data were filtered with a 10 Hz low-pass, zero-lag, 2^{nd} order Butterworth filter. Inverse dynamics were then computed to determine external moments for each joint (Gagnon & Gagnon, 1992), which were then normalized to each participant's height (meters) and mass (kilograms). Kinematic and kinetic data were then exported to MATLAB (MathWorks, Inc., Natwick, MA) for further reduction using custom-written script.

For reduction of MVIC and EMG data, the peak force (N) for each MVIC condition was multiplied by the moment arm length (as determined by Dempster's data and accounting for placement of the dynamometer) (Dempster, 1955), then divided by participant body mass, resulting in a normalized torque, $N \cdot m \cdot kg^{-1}$. These torque values were used to represent maximum torque generation. To obtain MVIC muscle activation amplitude, the EMG data were filtered in MATLAB using a band-pass 20-350 Hz, 4^{th} order, zero-lag, Butterworth filter with full-wave rectification, and was processed using a root mean squared (RMS) algorithm with a 25-millisecond time constant. The peak RMS EMG amplitude was extracted from each trial and averaged within each condition, resulting in an averaged peak EMG amplitude for each MVIC direction that was used to normalize the EMG signal (% MVIC) obtained during the landing task. Therefore, peak amplitude was defined as the averaged peak RMS signal across trials within each MVIC condition.

MATLAB script was written to extract all variables from the exported data. Specifically, initial ground contact (the point at which vertical ground reaction force exceeded 10N) marked the beginning of the landing phase. The end of the landing phase was the point of maximum knee flexion. Initial hip and knee angles (sagittal, frontal, and transverse) were defined as the respective joint angles at the moment of initial ground contact. Peak hip and knee angles were also extracted (sagittal, frontal, and transverse planes), and were defined as the maximum respective joint angle occurring during the landing phase. Joint excursions were calculated as the peak angle minus the initial angle, in degrees. Peak external moments were obtained for the hip and knee in each cardinal plane, and were defined as the maximum normalized moment during the landing phase.

Statistical Approach. Multiple linear regressions were used to determine the direct influence of femoral anteversion and hip ROM on functional valgus collapse and the mediating effect of gluteal strength and activation. Femoral anteversion and ROM_{IR}, and ROM_{ER} were entered on the first step of separate forward stepwise multiple linear regressions to predict frontal and transverse plane hip and knee 1) angles at initial contact, 2) peak angles, 3) joint excursions, and 4) peak joint moments (p entry < .20). The relative contributions of each independent variable were inspected using semi-partial (part) correlations. Strength and activation for each gluteal muscle were then entered on the second step of the forward stepwise method, and R squared changes were examined. To quantify any mediating effects of gluteal strength and activation on relationships between anatomy and biomechanics, the adjusted upsilon (υ_{adj}) statistic was calculated for each instance in which the addition of neuromuscular variables weakened or strengthened part correlations between anatomy and biomechanics. υ_{adj} is an effect size statistic used to assess the degree of mediation, and is defined as the variance in Y accounted for jointly by M (mediator) and X. (Lachowicz, Preacher, & Kelley, in press). As a "variance explained" measure and in following with R^2 effect sizes, recommended small, medium, and large effect sizes for υ_{adj} are .01, .15, and .25, respectively. The MBESS package (Kelley, 2017) in R (R Core Team, 2017) was used to compute υ_{adj} values. Lastly, because both empirical evidence and functional anatomy indicate that the amount of potential knee valgus is dependent upon sagittal plane kinematics (Fukuda et al., 2003), peak hip and knee flexion were included as control variables in all analyses. Furthermore, due to the potential sex-specific nature of this mechanism, all analyses were conducted separately for each sex. SPSS (Version 21, IBM Corp, Armonk, NY) was used for all analyses unless otherwise noted.

Results

Descriptive and bivariate correlation statistics for the independent variables are presented in Tables 5.1 and 5.2, respectively. As expected, females had greater femoral anteversion and ROM_{IR} and less hip extension and hip abduction peak torque, and less gluteus medius muscle activation, thus confirming the need for sex-stratified analyses (Table 5.1). Complete stepwise multiple regression results for frontal and transverse plane hip and knee biomechanics are presented in Tables 5.3-5.6. Results will be presented by joint and plane (e.g. frontal plane hip biomechanics). Complete results will be presented for the female cohort first, followed by results for the male cohort.

Table 5.1. Descriptive Statistics for Independent Variables in Males and Females.

	Fem. Ant. (°)	ROM_{IR} (°)	ROM_{ER} (°)	HEX_{PTQ} ($N \cdot m \cdot kg^{-1}$)	$HABD_{PTQ}$ ($N \cdot m \cdot kg^{-1}$)	$HEX_{\%MVIC}$ (% MVIC)	$HABD_{\%MVIC}$ (% MVIC)
Females	9.4±4.5	30.4±10.3	47.8±7.6	.86±.21	1.15±.24	37.2±31.4	42.2±23.7
Males	3.4±4.4	20.1±8.5	49.2±6.4	1.04±.26	1.49±.29	31.9±29.2	80.6±114.6
Totals	6.4±4.5*	25.3±9.4*	48.5±7.0	.95±.24*	1.32±.27*	34.6±30.3	61.4±69.2*

Fem. Ant. = femoral anteversion; ROM_{IR} = internal rotation ROM; ROM_{ER} = external rotation ROM; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; *significant difference at p<.05

Table 5.2. Within-Sex Bivariate Correlations Among Independent Variables.

	Fem. Ant.	ROM_{IR}	ROM_{ER}	HEX_{PTQ}	$HABD_{PTQ}$	$HEX_{\%MVIC}$	$HABD_{\%MVIC}$
Fem. Ant.	--	**.76***	**-.42***	**.15**	**-.26***	**-.00**	**-.21**
ROM_{IR}	.34*	--	**-.42***	**.27***	**-.28***	**-.28***	**-.39***
ROM_{ER}	-.31*	-.36*	--	**.01**	**.00**	**.25**	**.00**
HEX_{PTQ}	-.17	-.16	.11	--	**.58***	**-.33***	**-.05**
$HABD_{PTQ}$	.04	-.32*	-.09	.27*	--	**-.15**	**-.22**
$HEX_{\%MVIC}$	-.04	.22	-.10	-.12	.01	--	**.53***
$HABD_{\%MVIC}$	-.38*	.04	-.30*	.07	-.11	.57*	--

Bolded values signify female correlations; unbolded values signify male correlations; Fem. Ant. = femoral anteversion; ROM_{IR} = internal rotation ROM; ROM_{ER} = external rotation ROM; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; * significant correlation at p <.05

Frontal Plane Hip Biomechanics in Females. After controlling for sagittal plane hip and knee position, anatomical variables did not predict initial position, peak position, joint excursion, or joint moment. Neuromuscular variables significantly predicted peak frontal plane hip angles (Table 5.3).

Peak Frontal Plane Hip Angles. Anatomical variables did not predict peak frontal plane hip position in females. When neuromuscular variables were added, lesser hip abduction torque predicted greater peak hip adduction (part r = -.32; R^2 change =.10, p =.02), while greater hip extension torque predicted greater peak hip abduction (part r = -.29; R^2 change =.08, p =.05).

Table 5.3. Final GLM Forward Stepwise Regression Models Detailing the Influence of Control Variables, Anatomical Variables, and Neuromuscular Variables on *Frontal Plane Hip Biomechanics.*

			P Value		Part Correlations								
					Control Variables		Anatomical Variables			Neuromuscular Variables			
Dependent Variable	Step	Sex	R^2	R^2 change	Peak Knee Flexion	Peak Hip Flexion	Fem. Ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
Initial Hip Add	1	F	.16 (.02)		-.31	.27							
(+)	2		.16 (.02)	--	-.31	.27	--	--	--				
	3		.21 (.02)	.05 (.13)	-.27	.31	--	--	--	--	-.22	--	--
	1	M	.09 (.15)		.04	.28							
	2		.28 (.01)	.19 (.01)	-.02	.27	.20	--	.43				
	3		.31 (.01)	.04 (.15)	.01	.29	.16	--	.43	-.20	--	--	--
Peak Hip Add	1	F	.18 (.02)		-.22	.37							
(+)	2		.18 (.02)	--	-.22	.37	--	--	--				
	3		.28 (.003)	.10 (.02)	-.17	.42	--	--	--	--	-.32	--	--
	1	M	.01 (.74)		.09	.07							
	2		.27 (.02)	.26 (.01)	.10	-.03	.30	.28	.38				
	3		.27 (.02)	--	.10	-.03	.30	.28	.38	--	--	--	--
Peak Hip Abd	1	F	.13 (.05)		-.31	.19							
(-)	2		.13 (.05)	--	-.31	.19	--	--	--				
	3		.21 (.02)	.08 (.05)	-.26	.18	--	--	--	-.29	--	--	--
	1	M	.03 (.51)		.06	.16							
	2		.22 (.04)	.19 (.01)	.01	.13	.26	--	.41				
	3		.22 (.04)	--	.01	.13	.26	--	.41	--	--	--	--
Hip Add Exc	1	F	.04 (.44)		.20	.00							
(+)	2		.04 (.44)	--	.20	.00	--	--	--				
	3		.09 (.28)	.05 (.14)	.14	.00	--	--	--	--	--	--	.23
	1	M	.06 (.28)		.09	-.24							
	2		.26 (.01)	.21 (.01)	.17	-.35	.26	.28	--				
	3		.26 (.01)	--	.17	-.35	.26	.28	--	--	--	--	--
Hip Abd Exc	1	F	.10 (.11)		.13	-.29							

(-)	2		.10 (.11)	--	.13	-.29	--	--	--				
	3		.10 (.11)	--	.13	-.29	--	--	--	--	--	--	--
	1	M	.20 (.01)		.05	-.44							
	2		.25 (.01)	.05 (.10)	.07	-.49	.23	--	--				
	3		.25 (.01)	--	.07	-.49	.23	--	--	--	--	--	--
Hip Add Mom	1	F	.24 (.003)		.01	.49							
(+)	2		.31 (.002)	.06 (.06)	-.01	.50	--	.25	--				
	3		.33 (.002)	.03 (.20)	-.04	.46	--	.29	--	--	.17	--	--
	1	M	.00 (1.00)		-.01	.002							
	2		.26 (.01)	.26 (.002)	-.03	-.07	.44	--	.38				
	3		.30 (.01)	.04 (.15)	-.00	-.05	.40	--	.39	-.20	--	--	--
Hip Abd Mom	1	F	.09 (.14)		-.17	-.24							
(-)	2		.17 (.10)	.08 (.15)	-.20	-.25	.20	-.28	--				
	3		.21 (.09)	.04 (.16)	-.16	-.25	.17	-.22	--	-.21	--	--	--
	1	*M	.02 (.67)		-.01	.14							
	2		.02 (.67)	--	-.01	.14	--	--	--				
	3		.15 (.10)	.13 (.02)	-.07	.11	--	--	--	--	-.35	--	--

p entry<.20; Add = adduction; Abd = abduction; Exc = excursion; Mom = moment; Fem. Ant. = Femoral Anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension activation amplitude; HABD%MVIC = hip abduction activation amplitude; *N=44, case excluded as a multivariate outlier (Cook's D >1).

Transverse Plane Hip Biomechanics in Females. After controlling for sagittal plane hip and knee position, anatomical variables predicted initial hip rotation, peak hip rotation, and peak hip rotary moments. After accounting for the anatomical variables, the neuromuscular variables did not explain any additional variance in transverse plane hip biomechanics in females, nor did they provide any meaningful mediation effects (Table 5.4).

Initial Transverse Plane Hip Angle. Lesser femoral anteversion (part r = -.22) and greater ROM_{IR} (part r =.42) predicted greater initial hip internal rotation (R^2 change=.20, p=.01). The addition of neuromuscular variables did not significantly improve the overall model (R^2 change=.03, p=.19). While the addition of lesser gluteus maximus activation weakened the part correlations between anatomy (femoral anteversion Δ part r =.04, υ_{adj} < .01; ROM_{IR} Δ part r = -.10) and greater initial hip internal rotation angle, the mediation effect size was negligible (υ_{adj} < .01).

Peak Transverse Plane Hip Angles. Greater ROM_{IR} predicted greater hip internal rotation (part r =.35; R^2 change =.12, p =.02), while greater ROM_{IR} (part r =.46) and lesser femoral anteversion (part r = -.26) predicted lesser hip external rotation (R^2 change =.23, p =.004). The addition of neuromuscular variables did not significantly improve the overall model (R^2 change =.05, p =.11), though greater gluteus maximus activation did enter the model (part r = -.22). While this weakened the part correlations between anatomy (femoral anteversion Δ part r =.04, υ_{adj} < .01; ROM_{IR} Δ part r = -.12) and greater peak hip external rotation angle, the mediation effect size was negligible (υ_{adj} < .01).

Peak Transverse Plane Hip Joint Moments. Lesser ROM_{ER} (part r = -.35) predicted greater hip internal rotation moment (R^2 change =.12, p =.02). The addition of neuromuscular variables neither improved the overall model nor altered the part correlation between ROM_{ER} and hip internal rotation moment.

Table 5.4. Final GLM Forward Stepwise Regression Models Detailing the Influence of Control Variables, Anatomical Variables, and Neuromuscular Variables on *Transverse Plane Hip Biomechanics.*

			P Value		Part Correlations								
					Control Variables		Anatomical Variables			Neuromuscular Variables			
Dependent Variable	Step	Sex	R^2	R^2 change	Peak Knee Flexion	Peak Hip Flexion	Fem. Ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
Initial Hip IR	1	F	.08 (.19)		.02	.27							
(+)	2		.28 (.01)	.20 (.01)	.04	.29	-.22	.42	--				
	3		.31 (.01)	.03 (.19)	.01	.29	-.18	.32	--	--	--	-.18	--
	1	M	.00 (.97)		-.04	.01							
	2		.11 (.17)	.11 (.03)	.02	-.02	--	--	-.34				
	3		.11 (.17)	--	.02	-.02	--	--	-.34	--	--	--	--
Peak Hip IR	1	F	.11 (.10)		-.04	.32							
(+)	2		.23 (.01)	.12 (.02)	-.07	.33	--	.35	--				
	3		.23 (.01)	--	-.07	.33	--	.35	--	--	--	--	--
	1	M	.02 (.61)		-.15	-.01							
	2		.23 (.03)	.21 (.01)	-.07	-.10	.23	--	-.31				
	3		.23 (.03)	--	-.07	-.10	.23	--	-.31	--	--	--	--
Peak Hip ER	1	F	.05 (.35)		-.13	.18							
(-)	2		.27 (.01)	.23(.004)	-.09	.19	-.26	.46	--				
	3		.32 (.01)	.05 (.11)	-.13	.19	-.22	.34	--	--	--	-.22	--
	1	M	.02 (.71)		-.13	-.01							
	2		.22 (.04)	.21 (.01)	-.04	-.09	.19	--	-.34				
	3		.22 (.04)	--	-.04	-.09	.19	--	-.34	--	--	--	--
Hip IR Exc	1	*F	.09 (.14)		-.20	-.23							
(+)	2		.09 (.14)	--	-.20	-.23	--	--	--				
	3		.09 (.14)	--	-.20	-.23	--	--	--	--	--	--	--
	1	M	.04 (.45)		-.18	-.04							
	2		.04 (.45)	--	-.18	-.04	--	--	--				
	3		.13 (.22)	.09 (.13)	-.21	-.00	--	--	--	--	--	-.30	.21
Hip ER Exc	1	F	.18 (.02)		-.36	-.24							
(-)	2		.18 (.02)	--	-.36	-.24	--	--	--				

	3		.18 (.02)	--	-.36	-.24	--	--	--	--	--	--	--
	1	M	.04 (.48)		-.17	-.04							
	2		.04 (.48)	--	-.17	-.04	--	--	--				
	3		.04 (.48)	--	-.17	-.04	--	--	--	--	--	--	--
Hip IR Mom	1	F	.00 (.94)		.03	.04							
(+)	2		.13 (.14)	.12 (.02)	-.01	-.04	--	--	-.35				
	3		.18 (.08)	.06 (.10)	-.05	-.04	--	--	-.36	.24	--	--	--
	1	M	.02 (.72)		.10	.06							
	2		.20 (.06)	.18 (.02)	.14	-.01	--	.42	.20				
	3		.30 (.03)	.10 (.08)	.07	.04	--	.44	.17	.25	--	-.21	--
Hip ER Mom	1	F	.38(<.001)		-.28	-.54							
(-)	2		.38(<.001)	--	-.28	-.54	--	--	--				
	3		.38(<.001)	--	-.28	-.54	--	--	--	--	--	--	--
	1	M	.11 (.09)		-.30	-.08							
	2		.33 (.01)	.22 (.01)	-.24	-.08	-.34	.29	-.23				
	3		.47 (.001)	.14 (.01)	-.31	-.02	-.33	.31	-.25	.29	--	-.25	--

p entry<.20; IR = internal rotation; ER = external rotation; Exc = excursion; Mom = moment; Fem. Ant. = femoral anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension activation amplitude; HABD%MVIC = hip abduction activation amplitude; *N=44, case excluded as a multivariate outlier (Cook's D >1)

Frontal Plane Knee Biomechanics in Females. After controlling for sagittal plane hip and knee position, the anatomical variables were associated with peak frontal plane knee angle, knee joint excursion, and peak frontal plane knee moments (Table 5.5). The addition of neuromuscular variables did not explain additional variance, nor did it alter any relationships between anatomy and frontal plane knee biomechanics.

Frontal Plane Knee Angles. Greater ROM_{ER} predicted greater peak knee adduction (part r =.44; R^2 change =.20, p =.001), and greater knee adduction excursion (part r = .30; R^2 change =.09, p =.05).

Frontal Plane Knee Joint Moments. Greater ROM_{IR} (part r = .35) and greater ROM_{ER} (part r = .36) predicted greater peak knee adduction moment (R^2 change =.18, p =.01).

Table 5.5. Final GLM Forward Stepwise Regression Models Detailing the Influence of Control Variables, Anatomical Variables, and Neuromuscular Variables on *Frontal Plane Knee Biomechanics.*

					Part Correlations								
			P Value		Control Variables		Anatomical Variables			Neuromuscular Variables			
Dependent Variable	Step	Sex	R^2	R^2 change	Peak Knee Flexion	Peak Hip Flexion	Fem. Ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
Initial Knee Add	1	F	.23 (.004)		.43	-.22							
(+)	2		.28 (.003)	.05 (.10)	.46	-.17	--	--	.23				
	3		.28 (.003)	--	.46	-.17	--	--	.23	--	--	--	--
	1	M	.00 (.93)		.04	-.05							
	2		.26 (.02)	.26 (.07)	-.01	.03	--	-.50	-.26				
	3		.26 (.02)	--	-.01	.03	--	-.50	-.26	--	--	--	--
Peak Knee Add	1	F	.21 (.01)		.43	-.17							
(+)	2		.40(<.001)	.20 (.001)	.48	-.06	--	--	.44				
	3		.40(<.001)	--	.48	-.06	--	--	.44	--	--	--	--
	1	M	.00 (.93)		-.03	.05							
	2		.22 (.04)	.21 (.05)	-.07	.12	--	-.44	-.30				
	3		.22 (.04)	--	-.07	.12	--	-.44	-.30	--	--	--	--
Peak Knee Abd	1	F	.17 (.02)		.32	-.26							
(-)	2		.21 (.02)	.04 (.15)	.36	-.26	-.20	--	--				
	3		.21 (.02)	--	.36	-.26	-.20	--	--	--	--	--	--
	1	M	.00 (.93)		-.06	-.01							
	2		.25 (.02)	.25 (.01)	-.07	.05	--	-.45	-.36				
	3		.25 (.02)	--	-.07	.05	--	-.45	-.36	--	--	--	--
Knee Add Exc	1	F	.01 (.88)		.01	.08							
(+)	2		.09 (.26)	.09 (.05)	.05	.14	--	--	.30				
	3		.09 (.26)	--	.05	.14	--	--	.30	--	--	--	--
	1	M	.07 (.23)		-.16	.23							
	2		.07 (.23)	--	-.16	.23	--	--	--				
	3		.07 (.23)	--	-.16	.23	--	--	--	--	--	--	--
Knee Abd Exc	1	F	.06 (.31)		-.23	-.04							

(-)	2		.06 (.31)	--	-.23	-.04	--	--	--				
	3		.06 (.31)	--	-.23	-.04	--	--	--	--	--	--	--
	1	M	.05 (.37)		-.21	.08							
	2		.15 (.08)	.10 (.03)	-.16	.05	--	--	-.32				
	3		.15 (.08)	--	-.16	.05	--	--	-.32	--	--	--	--
Knee Add Mom	1	F	.16 (.03)		.32	.23							
(+)	2		.33 (.01)	.18 (.01)	.33	.32	--	.35	.36				
	3		.37 (.01)	.04 (.13)	.29	.28	--	.39	.38	--	.20	--	--
	1	M	.03 (.53)		.14	.08							
	2		.14 (.11)	.11 (.03)	.08	.15	--	-.33	--				
	3		.18 (.08)	.05 (.13)	.03	.14	--	-.38	--	--	-.22	--	--
Knee Abd Mom	1	F	.02 (.72)		-.02	-.12							
(-)	2		.02 (.72)	--	-.02	-.12	--	--	--				
	3		.07 (.36)	.06 (.11)	.02	-.12	--	--	--	-.24	--	--	--
	1	M	.05 (.36)		-.12	.20							
	2		.18 (.08)	.13 (.05)	-.12	.23	--	-.30	-.30				
	3		.28 (.02)	.09 (.03)	-.17	.23	--	-.40	-.36	--	-.31	--	--

p entry<.20; Add = adduction; Abd = abduction; Exc = Excursion; Mom = moment; Fem. Ant. = femoral anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension activation amplitude; HABD%MVIC = hip abduction activation amplitude

Transverse Plane Knee Biomechanics in Females. After controlling for sagittal plane hip and knee position, anatomical variables were associated with peak transverse plane knee angles. The addition of neuromuscular variables explained additional variance and altered part correlations between anatomy and transverse plane knee biomechanics (Table 5.6).

Initial Transverse Plane Knee Angle. Anatomical variables did not significantly predict initial knee position (R^2 change =.10, p =.12). Higher peak gluteus maximus activation predicted greater initial knee internal rotation (part r =.32; R^2 change =.10, p =.03) and while this strengthened the non-significant part correlation between greater ROM_{IR} (Δ part r =.09) and greater initial knee internal rotation angle, the mediation effect was small (υ_{adj} = .01).

Peak Transverse Plane Knee Angles. Greater ROM_{IR} (part r =.31) and ROM_{ER} (part r =.31), and lesser femoral anteversion (part r = -.18) predicted greater peak knee internal rotation (R^2 change =.16, p =.04). The inclusion of greater gluteus maximus activation explained additional variance (part r =.37; R^2 change =.14, p =.01), and while it strengthened the part correlations for ROM_{IR} and femoral anteversion (Δ part r = .11 and -.07), the mediation effects were small (υ_{adj} = .01 and < .01, respectively).

Table 5.6. Final GLM Forward Stepwise Regression Models Detailing the Influence of Control Variables, Anatomical Variables, and Neuromuscular Variables on *Transverse Plane Knee Biomechanics.*

			P Value		Part Correlations								
					Control Variables		Anatomical Variables			Neuromuscular Variables			
Dependent Variable	Step	Sex	R^2	R^2 change	Peak Knee Flexion	Peak Hip Flexion	Fem. Ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
Initial Knee IR	1	F	.01 (.82)		.08	.05							
(+)	2		.11 (.31)	.10 (.12)	.10	.13	--	.21	.30				
	3		.21 (.09)	.10 (.03)	.13	.13	--	.30	.28	--	--	.32	--
	1	M	.03 (.55)		.16	.02							
	2		.13 (.13)	.10 (.04)	.14	.10	-.31	--	--				
	3		.25 (.02)	.12 (.01)	.08	.06	-.29	--	--	--	-.35	--	--
Peak Knee IR	1	F	.12 (.07)		.34	-.05							
(+)	2		.28 (.02)	.16 (.04)	.40	.04	-.18	.31	.31				
	3		.42 (.001)	.14 (.01)	.44	.04	-.25	.42	.27	--	--	.37	--
	1	M	.11 (.08)		.30	.09							
	2		.22 (.02)	.11 (.02)	.27	.17	-.33	--	--				
	3		.39 (.001)	.17 (.01)	.18	.11	-.27	--	--	.16	-.41	--	--
Peak Knee ER	1	F	.00 (.92)		.06	.02							
(-)	2		.06 (.49)	.05 (.14)	.09	.07	--	--	.23				
	3		.12 (.25)	.07 (.09)	.12	.06	--	--	.16	--	--	.26	--
	1	M	.03 (.56)		.15	.04							
	2		.14 (.18)	.11 (.08)	.15	.07	-.32	.21	--				
	3		.26 (.03)	.12 (.02)	.08	.05	-.26	.06	--	--	-.34	--	--
Knee IR Exc	1	F	.18 (.02)		.40	-.16							
(+)	2		.18 (.02)	--	.40	-.16	--	--	--				
	3		.18 (.02)	--	.40	-.16	--	--	--	--	--	--	--
	1	M	.06 (.31)		.19	.10							
	2		.06 (.31)	--	.19	.10	--	--	--				
	3		.17 (.10)	.12 (.07)	.18	-.01	--	--	--	.20	--	--	.30
Knee ER Exc	1	F	.04 (.43)		-.13	-.15							

(-)	2		.04 (.43)	--	-.13	-.15	--	--	--				
	3		.04 (.43)	--	-.13	-.15	--	--	--	--	--	--	--
	1	M	.01 (.86)		-.08	.05							
	2		.09 (.29)	.08 (.07)	-.05	-.02	.28	--	--				
	3		.09 (.29)	--	-.05	-.02	.28	--	--	--	--	--	--
Knee IR Mom	1	F	.12 (.07)		.32	-.13							
(+)	2		.12 (.07)	--	.32	-.13	--	--	--				
	3		.12 (.07)	--	.32	-.13	--	--	--	--	--	--	--
	1	M	.06 (.30)		.18	.11							
	2		.06 (.30)	--	.18	.11	--	--	--				
	3		.06 (.30)	--	.18	.11	--	--	--	--	--	--	--
Knee ER Mom	1	F	.09 (.15)		.13	-.27							
(-)	2		.09 (.15)	--	.13	-.27	--	--	--				
	3		.15 (.07)	.07 (.08)	.07	-.27	--	--	--	--	--	--	.26
	1	M	.23 (.004)		.22	.39							
	2		.27 (.004)	.04 (.15)	.25	.34	--	.19	--				
	3		.27 (.004)	--	.25	.34	--	.19	--	--	--	--	--

p entry<.20; IR = internal rotation; ER = external rotation; Exc = excursion; Mom = moment; Fem. Ant. = femoral anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension activation amplitude; HABD%MVIC = hip abduction activation amplitude

Frontal Plane Hip Biomechanics in Males. After controlling for sagittal plane hip and knee position, anatomical variables were associated with initial and peak frontal plane hip position, frontal plane hip excursion, and frontal plane hip joint moments. While the inclusion of neuromuscular variables had no effect on frontal plane hip angles, they did explain additional variance in adduction moments without altering the part correlations with anatomy (Table 5.3).

Initial Frontal Plane Hip Angles. Greater femoral anteversion (part r = .20) and ROM_{ER} (part r = .43) predicted greater initial hip adduction (R^2 change =.19, p =.01).

Peak Frontal Plane Hip Angles. A combination of greater femoral anteversion (part r = .30), greater ROM_{IR} (part r =.28), and greater ROM_{ER} (part r = .38) predicted greater peak hip adduction angle (R^2 change =.26, p =.01). Similarly, greater femoral anteversion (part r = .26) and greater ROM_{ER} (part r = .41) predicted less peak hip abduction (R^2 change =.19, p =.01).

Frontal Plane Hip Joint Excursions. Greater femoral anteversion (part r = .26) and greater ROM_{IR} (part r = .28) predicted greater hip adduction excursion (R^2 change =.21, p =.01).

Frontal Plane Hip Joint Moments. Greater femoral anteversion (part r = .44) and ROM_{ER} (part r = .38) predicted greater hip adduction moment (R^2 change =.26, p =.002). The inclusion of neuromuscular variables neither significantly improved this model, nor did it alter the part correlations between femoral anteversion, ROM_{ER}, and hip adduction moment.

Although anatomy did not predict hip abduction moment, greater hip abduction torque did predict greater hip abduction moment (part r = -.35; R^2 change =.13, p =.02).

Transverse Plane Hip Biomechanics in Males. After controlling for sagittal plane hip and knee position, anatomical variables were associated with initial and peak hip angles, and peak hip joint moments in the transverse plane. The inclusion of neuromuscular variables strengthened the prediction of transverse plane hip biomechanics in some cases, but this did not alter the relationships between anatomy and transverse plane hip biomechanics (Table 5.4).

Initial Transverse Plane Hip Angle. Lesser ROM_{ER} (part r = -.34) predicted greater initial hip internal rotation (R^2 change =.11, p =.03). The addition of neuromuscular variables neither improved the overall model nor altered the part correlation between ROM_{ER} and initial hip internal rotation.

Peak Transverse Plane Hip Angles. Greater femoral anteversion and lesser ROM_{ER} predicted greater peak hip internal rotation (part r = .23 and -.31, respectively; R^2 change =.21, p =.01) and lesser peak hip external rotation (part r = .19 and -.34, respectively; R^2 change =.21, p =.01). The addition of neuromuscular variables neither improved the overall model nor altered any part correlations between anatomy and peak transverse plane hip angles.

Peak Transverse Plane Hip Joint Moments. Greater ROM_{IR} (part r = .42) and ROM_{ER} (part r = .20) predicted greater hip internal rotation moment (R^2 change =.18, p =.02). The addition of neuromuscular variables neither improved the overall model nor altered any part correlations between anatomy.

Greater femoral anteversion (part r = -.34), greater ROM_{ER} (part r = -.23), and lesser ROM_{IR} (part r = .29) predicted greater hip external rotation moment (R^2 change =.22, p =.01). The inclusion of lesser hip extension torque (part r = .29) and greater gluteus maximus activation (part r = -.25) explained additional variance in greater hip external rotation moment (R^2 change =.14, p =.01), but this did not affect the anatomical part correlations (femoral anteversion Δ part r =.01, υ_{adj} < .01; ROM_{IR} Δ part r = .02; ROM_{ER} Δ part r = -.02), as the mediation effect sizes were negligible (υ_{adj} < .01).

Frontal Plane Knee Biomechanics in Males. After controlling for sagittal plane hip and knee position, anatomical variables were associated with peak knee angle, knee joint excursions, and peak knee joint moments in the frontal plane. The inclusion of neuromuscular variables both strengthened the prediction of frontal plane knee biomechanics and altered the relationships with anatomy (Table 5.5).

Peak Frontal Plane Knee Angle. Lesser ROM_{IR} and ROM_{ER} predicted greater peak knee adduction (part r = -.44 and -.30, respectively; R^2 change =.21, p =.05) and lesser peak knee abduction (part r = -.45 and -.36, respectively; R^2 change =.25, p =.01). The addition of neuromuscular variables neither improved the overall model nor altered any part correlations between anatomy and peak frontal plane knee angles.

Frontal Plane Knee Joint Excursions. Greater ROM_{ER} predicted greater knee abduction excursion (part r = -.32; R^2 change =.10, p =.03). The addition of neuromuscular variables neither improved the overall model nor altered the part correlations between ROM_{ER} and frontal plane knee excursion.

Peak Frontal Plane Knee Joint Moments. Lesser ROM_{IR} (part r = -.33) predicted greater knee adduction moment (R^2 change =.11, p =.03). Greater ROM_{IR} (part r = -.30) and greater ROM_{ER} (part r = -.30) predicted greater knee abduction moment (R^2 change =.13, p =.05). The inclusion of greater hip abduction torque further predicted greater knee abduction moment (part r = -.31; R^2 change =.09, p =.03), which resulted in stronger part correlations for ROM_{IR} (Δ part r = -.10, υ_{adj} < .01) and ROM_{ER} (Δ part r = -.06, υ_{adj} < .01).

Transverse Plane Knee Biomechanics in Males. After controlling for sagittal plane hip and knee position, anatomical variables were associated with initial and peak transverse plane knee angle. The inclusion of neuromuscular variables strengthened the overall prediction of transverse plane knee biomechanics, while weakening the part correlations between anatomy and transverse plane knee biomechanics (Table 5.6).

Initial Transverse Plane Knee Angle. Lesser femoral anteversion (part r = -.31) predicted greater initial knee internal rotation (R^2 change =.10, p =.04). The inclusion of lesser hip abduction torque further predicted greater initial knee internal rotation (part r = -.35; R^2 change =.12, p =.01), while the part correlation for femoral anteversion remained constant (Δ part r = .02), resulting in a negligible mediation effect (υ_{adj} < .01).

Peak Transverse Plane Knee Angles. Lesser femoral anteversion predicted greater peak knee internal rotation (part r = -.33; R^2 change =.11, p =.02), while lesser femoral anteversion (part r = -.32) and greater ROM_{IR} (part r = .21) predicted lesser peak knee external rotation (R^2 change =.12, p =.02). After controlling for anatomical variables, greater hip extension torque (part r = .16) and lesser hip abduction torque (part r = -.41) further predicted greater knee internal rotation (R^2 change =.17, p =.01), while lesser hip abduction torque predicted lesser knee external rotation (part r = -.34; R^2 change =.12, p =.02). After the addition of neuromuscular variables to the models, femoral anteversion's part correlations weakened (peak knee internal rotation Δ r =.06, υ_{adj} < .01; peak knee external rotation Δ r = .06), as did the part correlation between ROM_{IR} and peak knee external rotation (Δ r = -.15). The computed mediation effect size, however, was small to negligible (υ_{adj} < .01 and υ_{adj} = .01, respectively).

Discussion

The identification of modifiable risk factors is paramount for the prevention of high risk biomechanics, and thus ACL injuries. Because one's anatomy is not readily modifiable, gluteal control of hip and knee motion may represent a viable target for intervention if that control has the ability to mediate (i.e. lessen) the effects of the anatomical variables. Previous research on the role of gluteal function in controlling lower extremity movement has been inconclusive (Cashman, 2012; Cronström, Creaby, Nae, & Ageberg, 2016b; Howard, Fazio, Mattacola, et al., 2011; Kaneko & Sakuraba, 2013a; Sigward et al., 2008). However, to our knowledge this is the first study to account for the shared variance between femoral alignment (both bony and capsular) and gluteal function in its prediction of functional valgus collapse. Together these variables tell a more complete story, given that femoral alignment could alter gluteal muscle length, thus impacting its functionality. This is also one of few studies to conduct sex specific analyses, which may yield potential sex-specific associations with respect to high-risk biomechanical strategies (Quatman & Hewett, 2009).

We hypothesized that an anatomical bias towards internal hip rotation, as evidenced by greater femoral anteversion, greater ROM_{IR}, and lesser ROM_{ER}, would precipitate greater movements and moments that contribute to functional valgus collapse. We also expected that strength and activation of the gluteus maximus and medius would explain additional variance in these biomechanical variables and that inefficient use of the gluteus maximus and gluteus medius, operationally defined as gluteal muscles displaying lower torque production and higher percentage activation, would mediate the relationships between femoral alignment and biomechanics. Our results partially supported these hypotheses. In males, greater femoral anteversion and greater ROM_{IR} and ROM_{ER} were associated with more functional valgus collapse, while in females, greater ROM_{IR} and *less* ROM_{ER} predicted riskier biomechanics. In both sexes, weaker gluteals activating to a higher degree explained additional biomechanical variance without having much effect on the correlations between anatomy and biomechanics, suggesting that the gluteal muscles may act independently of one's femoral alignment, and therefore might represent effective targets for injury prevention programs.

Gluteal Influences on Hip and Knee Biomechanics. In females, once accounting for the influence of femoral alignment on lower extremity biomechanics, gluteal function explained up to 14% of additional variance in hip and knee motion. Specifically, weaker hip extensors and hip abductors predicted smaller hip abduction (R^2 change = .08) but greater hip adduction (R^2 change = .10) peak angles. Furthermore, greater gluteus maximus activation predicted greater initial (R^2 change = .10) and peak (R^2 change = .14) knee internal rotation. These effects were consistent with our hypothesis that weaker muscles activating to a higher degree would predict greater functional valgus collapse, as hip adduction and knee internal rotation are primary components of a valgus collapse mechanism (Fukuda et al., 2003; Ireland, 1999). The additional variance explained by gluteal function without impacting the correlations with anatomy is important because it suggests that hip musculature may act independently of femoral alignment. As such, gluteal function may be modifiable regardless of one's bony and capsular structure. Generally, these data indicate that females with weak hip extensors and abductors exhibit greater hip adduction and knee internal rotation during a single-leg forward landing task. However, it is important to note that hip adduction is measured in relation to the pelvis, while knee internal rotation is measured relative to the femur. In a double-leg stance, the combination of hip adduction and tibial internal rotation (to the exclusion of hip internal rotation) could be difficult to explain. However, during a single-limb stance, if the contralateral pelvis was to drop (i.e. Trendelenberg's sign) and rotate away from the stance limb, this would increase relative hip adduction and external rotation, thus leaving the tibia in relative internal rotation. To confirm this theory we conducted a pair of posthoc regressions. Peak hip adduction (part r = -.30) and peak knee internal rotation (part r = -.30) combined to predict greater contralateral trunk rotation (R^2 = .14, p =.04), while greater peak ankle abduction was strongly correlated with peak knee internal rotation (R^2 = .66, p <.001). This suggests that females may compensate for weak gluteal muscles by contralaterally rotating their trunk during a single-leg land, which then increases hip adduction and tibial internal rotation.

In males, once accounting for the influences of femoral alignment on hip and knee biomechanics, gluteal function explained up to an additional 17% of the variance in biomechanical variables. Specifically, lower hip abduction strength predicted less hip and knee abduction moment, greater initial and peak knee internal rotation, and lesser peak knee external rotation. Additionally, the combination of less hip extension strength and greater gluteus maximus activation predicted greater hip external rotation moment. While the isolated effects on knee internal rotation are similar in males and females, the general influences of gluteal function between sexes are markedly different. Firstly, gluteal function in males was associated with joint moments (hip abduction, knee abduction, and hip external rotation), in contrast to females where observed effects were limited to kinematics. This is interesting because it suggests that males are able to use their musculature to generate torque about a joint. Females, on the other hand, who have weaker overall strength values when normalized to body weight, may not be able to use their gluteal muscles for torque generation, thus demonstrating another potential example of neuromuscular ineffectiveness. Secondly, while our data suggested that females used trunk motion to compensate for inefficient gluteal function, these patterns were not apparent in the male cohort. In particular, weaker hip abductors in males predicted smaller hip abduction and knee abduction joint moments. A common frontal plane compensation for a weak gluteus medius in females is an ipsilateral trunk lean, which in theory increases both hip and knee external abduction moments (Timothy E Hewett & Myer, 2011). However, this is opposite to what we observed in our male cohort, suggesting that weaker males do not compensate with an ipsilateral trunk lean, but are able to keep their center of mass over the stance limb, thus limiting hip abduction moment and maintaining a varus knee joint moment. Therefore, it's possible that males are able to decrease demands on weak hip abductors through alternative means, such as more exact foot placement, improved position sense, or preparatory neuromuscular activity.

The Mediating Effects of Gluteal Muscles on the Relationship between Femoral Alignment and Functional Valgus Collapse. While we hypothesized that gluteal function would mediate the relationships between one's femoral alignment and lower extremity biomechanics, this effect was not observed in either sex. In other words, the inclusion of gluteal muscle strength and activation did not meaningfully alter (neither weakened nor strengthened) the relationships between anatomy and biomechanics. To assess mediation, we employed the recently developed upsilon (υ) statistic (Lachowicz, Preacher, & Kelley, in press), which is defined as the percentage of variance in Y that is jointly explained by X and the mediating variable, while also controlling for spurious correlation arising from the ordering of variables. Recommended benchmarks for small, medium, and large effect sizes are .01, .15, and .25, respectively (Lachowicz, Preacher, & Kelley, in press). Our largest observed effect size was .01, while the majority of observed effect sizes were < .01. Even though this was contrary to our hypothesis, it still may be encouraging for the prospects of utilizing gluteal function to alter hip and knee biomechanics. If we had observed meaningful mediation, it would have demonstrated that gluteal function directly counteracts one's femoral alignment, and as such, gluteal function may have been mechanically linked to transverse plane femur alignment, bringing into question its ability to be modified. However, in our cohort, zero-order correlations between femoral alignment and gluteal function were low (*r* range = .00 - .39), indicating that altered placement of the greater trochanter may not incapacitate the gluteus medius and gluteus maximus. Therefore, it is possible that the gluteal muscles are able to affect safer movement patterns regardless of one's femoral alignment, though further research is needed to determine the efficacy of neuromuscular interventions in individuals with differing degrees of femoral alignment.

The Relationship between Femoral Alignment and Functional Valgus Collapse. The most consistent theme observed with respect to femoral alignment and lower extremity movement is that individuals with more lax hips, as indicated by greater ROM_{IR} and ROM_{ER}, were more likely to display motions and moments that contribute to functional valgus collapse. While this was true to some degree in both sexes, these relationships in the male cohort were limited to frontal plane hip and knee movement, whereas the effects in females were observed in both the frontal and transverse planes. In males, combinations of greater femoral anteversion with greater ROM_{IR} and greater ROM_{ER} were associated with more hip adduction and knee abduction. Thus, greater femoral anteversion with "looser" hips could be indicative of poor frontal plane hip and knee control in males. In females, more mobile hips were more strongly associated with transverse plane hip and knee movement. That is, combinations of greater ROM_{IR} and ROM_{ER} and less femoral anteversion were associated with greater hip and knee internal rotation movement. Similarly, combinations of greater ROM_{ER} and greater ROM_{IR} predicted greater initial and peak knee adduction angles and greater knee adduction moments. Generally, we observed greater ROM_{ER} alone to be more associated with safer biomechanics, while higher degrees of ROM_{IR} alone to be predictive of more risky biomechanics. Further research should examine the importance of the amount of ROM_{IR} relative to ROM_{ER} versus absolute ROM values. Excessive ROM_{IR} in the presence of minimal ROM_{ER} could be just as problematic, if not more problematic, than excessive ROM_{IR} and ROM_{ER}. If this is the case, then it may present an opportunity to intervene upon females' movement strategies. If it is possible to increase ROM_{ER} in females through rehabilitation, it may aid in the mitigation of potentially injurious frontal and transverse plane biomechanics.

Also of note, femoral anteversion displayed opposite effects in males and females. In males, greater femoral anteversion was indicative of more risky frontal plane hip movement, while in females lesser femoral anteversion was more likely to be associated with risky transverse plane hip and knee movement. However, ROM_{IR} and femoral anteversion in females were highly correlated (r = .76), suggesting that the influence of lesser femoral anteversion could be a spurious relationship. In males, ROM_{IR} and femoral anteversion appear to be separate constructs (r = .34), each contributing a unique portion of variance to movement patterns. Further research needs to be completed to determine specific influences of ROM_{IR} and the degree to which it is modifiable in males and females.

Limitations. We acknowledge limitations in our study. Although our sample of young, active individuals was one in which ACL injuries typically occur, it is possible that our included participants represented a survivor population. As such, our participants possibly had no need to compensate for sub-optimal femoral alignment, which could have resulted in a dampened effect.

We also acknowledge the assumption that peak joint angles represented the time at which participants were most vulnerable to injury. While this is common practice in lower extremity biomechanical research, the use of discrete peak joint angles may not have fully captured one's biomechanical risk profile. We also assumed that participants put forth maximal exertion during MVIC measurement. Though we did provide practice trials with verbal feedback, we cannot guarantee that obtained MVICs truly represented maximal effort.

Conclusion

In conclusion, this study demonstrated that in females, after accounting for greater ROM_{IR} and less ROM_{ER}, less gluteal strength and higher muscle activation was associated with greater functional valgus collapse. In males, after controlling for greater femoral anteversion and greater ROM_{IR} and ROM_{ER}, less gluteal strength and higher activation was associated with decreased joint moments consistent with a varus posture. In both sexes, the inclusion of femoral alignment and gluteal function was useful for predicting greater variance in biomechanical outcomes, despite the observation that gluteal function did not directly mediate the relationship between femoral alignment and functional valgus collapse. Further work is needed to determine whether passive ROM is a modifiable characteristic and whether interventions targeting gluteal function are effective in males and females with varying degrees of femoral alignment.

CHAPTER VI

MANUSCRIPT III. A PRELIMINARY MULTIVARIATE APPROACH TO ASSESS THE IMPACT OF GLUTEAL STRENGTH AND ACTIVATION ON FUNCTIONAL VALGUS COLLAPSE DURING A SINGLE-LEG FORWARD HOP LANDING

Abstract

Context. Functional valgus collapse, a coupled motion between hip adduction and internal rotation and knee valgus and rotation, is thought to be a contributor to ACL injury risk. Gluteal muscle strength and activation have been suggested to influence the degree of hip control, thus functional valgus collapse. However, the evidence supporting this role is mixed, possibly because findings to date are based on the limited assessment of initial and peak joint angles of individual motions, which are not necessarily timed relative to one another or with the point in the landing when the knee is most vulnerable. A more comprehensive assessment of integrated hip and knee motion across the entire landing phase may better elucidate the specific role of gluteal function.

Objective. Examine the roles of gluteus maximus and gluteus medius strength and activation on a 4-component linear combination of functional valgus collapse (hip adduction and internal rotation, knee abduction and internal rotation) *throughout* the landing phase of a single-leg forward hop, while controlling for hip anatomy (femoral anteversion, passive internal and external rotation ROM). We hypothesized that once accounting for hip anatomy, lower gluteal strength (MVIC) and higher gluteal activation (%MVIC) would be associated with a linear combination of greater functional valgus collapse (specifically the components of greater hip

internal rotation and adduction), and that these relationships would be observed early in the landing phase.

Design. Cross-sectional.

Setting. Research laboratory.

Patients or Other Participants. Forty-five females (20.1±1.7 yrs, 165.2±7.6 cm, 68.6±13.1 kg) and forty-five males (20.7±2.0 yrs, 177.7±8.5 cm, 82.8±16.3 kg) aged 18-25, with no history of lower extremity surgery and no injury history in the previous six months.

Intervention(s). Femoral anteversion and passive hip ROM were measured prone with the knee at 90°. Maximal voluntary isometric contractions were obtained for hip extension and abduction. Three-dimensional biomechanics and surface electromyography were obtained during performance of five trials of a single-leg forward landing task.

Main Outcome Measures. Statistical Parametric Mapping canonical correlation analyses (CCA) were used to identify time points throughout the landing phase in which there were significant associations between individual predictors (femoral alignment variables and neuromuscular variables) and a 4-component linear combination representing functional valgus collapse. Where significant associations were observed, General Linear Model (GLM) canonical correlations were then used to determine in the relevant time intervals the specific individual components of functional valgus where these associations were most prominent. Male and female cohorts were analyzed separately.

Results. In females, greater ROM_{ER} ($\beta_{st} = .46$) was associated with greater external hip adduction moment ($R^2 = .30$, $p = .01$) from 7-8% of the landing phase, then greater gluteus medius activation ($\beta_{st} = .50$) was associated with greater external hip adduction moment ($R^2 = .34$, $p = .06$) from 18% and 20% of the landing phase. In males, at 0% and from 2-3% of the landing phase, greater ROM_{IR} ($\beta_{st} = -.53$) and greater ROM_{ER} ($\beta_{st} = -.32$ and .52, respectively) were associated with greater knee abduction angle ($R^2 = .31$, $p = .01$) and greater hip adduction angle ($R^2 = .29$, $p = .02$). From 3-9% of the landing phase, greater hip extension peak torque ($\beta_{st} = -.37$ and .34, respectively) combined with greater ROM_{IR} ($\beta_{st} = -.52$ and -.33, respectively) to associate with greater knee abduction moment ($R^2 = .42$, $p = .01$) and lower hip adduction moment ($R^2 = .54$, $p = .001$).

Conclusions. In both the male and female cohorts, once accounting for anatomy, gluteal strength and activation was associated with hip and knee motion early in the landing phase. These neuromuscular associations were observed ~10 – 20ms after the onset of anatomical influences, and did not alter movement trajectories. Thus, it appears that anatomy may initially drive these motions, and that the gluteal muscles are not able to make appreciable corrections. Additionally, greater ROM_{IR} and ROM_{ER}, whether together or separately, appear to be the primary anatomical variables that promoted risky frontal plane hip and knee movement in the first few milliseconds of the landing phase.

Key Words. ACL, hip ROM, femoral anteversion, gluteal activation, gluteal strength, valgus, statistical parametric mapping

Introduction

Functional valgus collapse is thought to contribute to anterior cruciate ligament (ACL) ruptures (Timothy E Hewett et al., 2005; OKane et al., 2016), particularly in females (B P Boden et al., 2009; T E Hewett et al., 2009; T Krosshaug et al., 2007). Functional valgus collapse consists of coupled motion between the hip and knee (Hollman et al., 2014; Imwalle, Myer, Ford, & Hewett, 2009), that includes hip adduction and internal rotation, and knee abduction and internal rotation (Ireland, 1999; Meyer & Haut, 2008). Because of the dual-joint nature of the valgus collapse mechanism, much work has been dedicated to identifying risk factors which may influence one's ability to effectively control the hip, thus the knee.

As the primary external rotators and abductors of the hip, the gluteus maximus and gluteus medius muscles work eccentrically to control dynamic hip internal rotation and adduction (Starkey & Ryan, 2003), two primary components of functional valgus collapse. This is particularly true in a single-leg stance, when the gluteal muscles are tasked with maintaining a level pelvis (Starkey & Ryan, 2003). As such, strength and muscle activation of the gluteus maximus and gluteus medius have been suggested to impact hip control during functional movement (Howard, Fazio, Mattacola, Uhl, & Jacobs, 2011; Kaneko & Sakuraba, 2013; Sigward, Ota, & Powers, 2008), and therefore may represent an avenue for intervention and injury prevention. Specifically, some have suggested that decreased torque generation (MVIC) coupled with increased EMG amplitude is problematic and signifies an inefficient muscle (Homan et al., 2013a; A.-D. Nguyen et al., 2011). This neuromuscular profile would be suggestive of an individual with less effective neural drive, who would require a greater proportion of their strength to complete a given task. However, this observation is not consistent across all studies. Though both cross-sectional and prospective studies report significant associations between decreased hip strength with faulty biomechanics and increased ACL injury risk (Howard, Fazio, Mattacola, et al., 2011; Jacobs et al., 2007; Khayambashi et al., 2016b), systematic reviews attempting to link hip extensor and abductor function to lower extremity biomechanics have failed to demonstrate this consensus (Cashman, 2012; Cronström et al., 2016b; Rafeeuddin et al., 2016). One possible explanation for this lack of consensus could be the methods in which gluteal function is commonly examined. For example, demanding single-leg tasks have proven more effective in identifying differences in muscle function between groups (e.g. sex differences) than studies using less challenging double-leg tasks (Hollman et al., 2009b; Homan et al., 2013b; Howard, Fazio, Mattacola, et al., 2011; Jacobs et al., 2007; Lawrence et al., 2008; A.-D. Nguyen et al., 2011; Sigward et al., 2008; Thijs et al., 2007; Weinhandl et al., 2015; Willson et al., 2006). It has also been demonstrated that the moment arm of the gluteal muscles change as the hip moves into flexion (Delp et al., 1999), suggesting that gluteal muscle function could also shift throughout the landing phase as hip angle changes. As such, the contributions of the gluteal muscles may not be entirely captured by limiting analyses to peak angular joint positions. There is also reason to believe that the gluteal moment arm, thus function, may be impacted by one's hip anatomy. Specifically, a more internally rotated femur, as evidenced by greater femoral anteversion, greater internal rotation ROM, and lesser external rotation ROM may excessively lengthen the moment arm, thus reducing the effectiveness of the gluteal muscles in controlling lower extremity motion (Howard, Fazio, Mattacola, et al., 2011; Kaneko & Sakuraba, 2013a; J Nyland et al., 2004). Thus, examining gluteal function while also accounting for hip flexion angle and anatomy may aid in clarifying the role of the gluteal muscles in lower extremity control.

The lack of consensus surrounding associations between gluteal function and lower extremity control may also result from the common practice of analyzing only discrete variables (e.g. peak and initial angles, peak moments) for each individual joint motion. Because the hip and knee display frontal and transverse plane coupling (Hollman et al., 2014; Imwalle, Myer, Ford, & Hewett, 2009), it may not be sufficient nor accurate to analyze individual planes within each joint and make inferences to the entire movement pattern. While common, extracting only the initial and peak position variables from biomechanical time-series data makes the assumption that movement and loading are linear and that relative joint and planar contributions to overall lower extremity movement remain constant over time. Furthermore, when these peaks occur are not necessarily specific to the time frame most relevant to ACL injury, which are reported to occur 30 – 50ms after initial ground contact (Carlson et al., 2016; T Krosshaug et al., 2007; Tron Krosshaug et al., 2007), as peak forces are reported to occur 30 – 50ms after initial ground contact (T Krosshaug et al., 2007). Statistical Parametric Mapping (SPM) is a statistical method which allows for the analysis of complete time-series curves and for the inclusion of multiple dependent variables by creating a linear combination at each time point throughout the landing phase, resulting in a single variable which changes over time. SPM circumvents the conventional need to stringently adjust for Type I error rate, and instead accounts for the dependency of adjacent time points in its calculation of the appropriate significance threshold (Pataky, 2012). As such, this method may advance our understanding of gluteal muscle contributions to lower extremity control by examining time dependent associations between muscle strength and activation with the inter-related biomechanical variables across the entire landing phase.

Therefore, the objective of this exploratory study was to examine the roles of gluteal muscle strength and activation using a 4-component linear combination of functional valgus collapse (hip adduction and internal rotation, knee abduction and internal rotation) *during the entire* landing phase of a single-leg forward hop, while also accounting for the sagittal plane hip and knee position at landing and one's hip anatomy (femoral anteversion, passive internal and external rotation ROM). Where significant associations were observed, we then examined within each significant time frame the associations between gluteal muscle strength and activation and femoral alignment with functional valgus collapse and its components. We hypothesized that lower gluteal strength (MVIC) and higher gluteal activation (%MVIC) would be associated with the combined motions of greater functional valgus collapse early in the landing phase, and that these associations would be strongest for the hip components of greater hip adduction and internal rotation.

Methods

Participants. Forty-five female participants (20.1 ± 1.7 yrs, 165.2 ± 7.6 cm, 68.6 ± 13.1 kg) and forty-five male participants (20.7 ± 2.0 yrs, 177.7 ± 8.5 cm, 82.8 ± 16.3 kg) were recruited for a single session. This was part of a larger study powered to examine relationships at discrete time points between gluteal function and individual biomechanical variables, which accounted for five control variables, up to four independent variables, and one dependent variable. Thus, the current study was an exploratory analysis intended to build upon the original findings using a more integrated approach. To ensure a homogenous sample, specific inclusion criteria were 1) adults between the ages of 18 and 25 and 2) a score of two or more (at least "one time in a week") on categories 2-4 ("cutting, "decelerating", and "pivoting") of the Marx activity rating scale (see Appendix). Specific exclusion criteria were 1) any history of knee surgery, 2) any history of ligamentous or meniscal knee injury, 3) any history of lower extremity injury within the previous 6 months, 4) history or diagnosis of a vestibular condition affecting balance, and 5) history or diagnosis of any cardiovascular condition precluding exercise. Each participant provided written informed consent as approved by the university's Institutional Review Board. After obtaining written informed consent, each participant was asked to complete the following intake questionnaires: Physical Activity and Health History Questionnaire, Knee Outcome Survey (both the Activities of Daily Living Scale and the Sports Activities Scale), and the Marx Activity Rating Scale (Appendix).

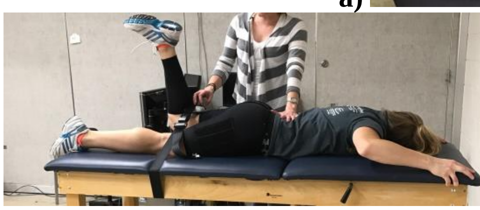
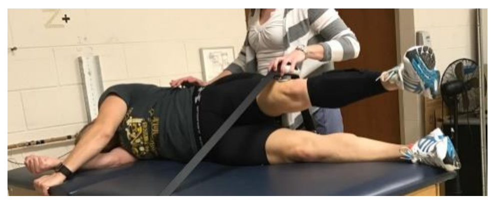
Anatomical Measures. Both femoral anteversion and hip ROM were measured prone with a standard inclinometer with the hip in neutral and the knee flexed to 90° (Magee, 1997). To measure femoral anteversion, the examiner rotated the lower leg while palpating the greater trochanter. When the greater trochanter was at its most lateral point, the transverse plane angle formed by the tibial shaft and true vertical was measured as femoral anteversion. To measure hip ROM, the examiner passively internally and externally rotated the lower leg while palpating the sacrum. At the point of initial sacral movement, the transverse angle formed by the tibial shaft and true vertical was measured as ROM_{IR} and ROM_{ER}, respectively (Figure 6.1a). Three trials were taken for each measure and averaged for analysis. The same examiner took all measurements, and had previously established good to excellent inter-day reliability [$ICC_{2,3}$(SEM); femoral anteversion: .92(1.2°); internal rotation ROM: .97(1.6°); external rotation ROM: .85(3.3°)] (Hogg et al, in press).

Surface Electromyography Instrumentation. Surface electromyography (EMG) signals were acquired with double differential electrodes (Trigno Wireless Sensors, Delsys, Boston, MA) from the gluteus medius and gluteus maximus during MVIC strength testing and performance of the single-leg forward landings. Prior to sensor placement, the skin was cleaned with an alcohol swab. The gluteus medius electrode was placed one-third the distance from the greater trochanter to the iliac crest (Rainoldi et al., 2004). The electrode on the gluteus maximus was placed one-third the distance from the second sacral vertebrae to the greater trochanter (Rainoldi et al., 2004). All electrodes were positioned parallel to muscle fiber orientation and secured with tape or prewrap. Proper positioning was verified with manual muscle testing and visual inspection of the EMG signal using EMGWorks (Delsys, Boston, MA). EMG data were sampled at 1000Hz and were manually synced with kinematic and kinetic data during post-collection processing.

Maximal Voluntary Isometric Contractions (MVICs). Prior to obtaining MVICs, each participant completed a five-minute warm up on a stationary bike at a self-selected pace. Following warm-up, MVICs of the gluteus maximus (hip extension) and gluteus medius (hip abduction) were obtained and used as representations of maximum torque generation values, as well as for normalization of EMG amplitude. For all MVIC measures, a strap was used to secure the dynamometer in place and to provide resistance for the participant. For MVIC measurement of the hip extensors, the participant was positioned prone with the knee bent to 90 degrees. With a handheld dynamometer (Lafayette Instruments, Boston, MA) placed over the posterior distal thigh two inches proximal to the joint line, the participant was asked to maximally contract into hip extension (Bohannon, 1986; Starkey & Ryan, 2003). Hip abduction MVICs were measured side-lying on the right side, with the left leg up. The left leg was placed in 10-15 degrees of hip extension and slightly externally rotated, thus isolating the gluteus medius. Maximal hip abduction was resisted by placing the lower edge of the dynamometer two inches proximal to the lateral knee joint line (Krause et al., 2007). Prior to collecting each MVIC, participants were familiarized to the measure and allowed up to three submaximal practice trials. Each MVIC condition consisted of three 5-second trials, with 30 seconds rest between trials. To prevent an artificial spike in dynamometer output during collection, each participant was instructed to slowly increase force output, reaching maximum force production during the third second of the five second trial. Prior to data collection the PI established reliability for strap-assisted handheld dynamometry [$ICC_{2,2}$(SEM); hip extension: .76(3.4kg); hip abduction: .96(1.6kg)].

Figure 6.1. Measurement Position of Anatomical Variables (a), Hip Extension MVIC (b), and Hip Abduction MVIC (c).

a) 

b)  **c)** 

Procedure for Single-Leg Forward Landing. Participants were familiarized to the task prior to data collection and were allowed up to three practice trials. Tape was placed on the ground at a distance equal to 40% of each participant’s height away from the front edge of the forceplate. A foam barrier equal to 15% of the participant’s height (Jacobs et al., 2007) was placed halfway between the tape and the edge of the forceplate. Instructions to participants were standardized. Participants were instructed to stand behind the tape and jump from 2 legs over the foam barrier, landing on their left foot (Figure 6.2). Trials were discarded if the participant double-hopped upon landing, hit the barrier, didn’t clear the barrier with both feet, didn’t land with the entire left foot on the forceplate, or used their contralateral limb for additional support. Five clean trials were collected and used for analysis.

Figure 6.2. Terminal Position for the Single-Leg Forward Landing Task.

Biomechanical Instrumentation. Prior to digitization for motion capture, participants were outfitted with standardized shoes (Adidas Uraha 2, Adidas AG, Herzogenaurach, Bavaria) to eliminate potential shoe-surface interactions. Participants were adorned with six marker clusters, placed at each of the following locations: lateral aspect of the left foot, lateral aspect of the left lower leg (mid-shaft), medial and lateral proximal tibial flares, the lateral left thigh (mid-shaft), the L5-S1 junction, and the postero-superior thorax (C7-T1 spinous processes). Participants were then digitized using the MotionMonitor software (Innovative Sports Training, Chicago, IL). Joint centers for the knee and ankle were defined as the midway point between the medial and lateral femoral condyles and medial and lateral malleoli, respectively. The hip joint center was determined using the Bell method (A. L. Bell & Pedersen, 1989). A segment-based coordinate system was used to define each body segment. The X-axis was defined as the anterior-posterior axis (adduction/abduction), the Y-axis was the distal-proximal axial axis (internal/external rotation), and the Z-axis was defined as the medial-lateral axis (flexion/extension). Motions for each joint were calculated using Euler's equations (Z Y' X") (Kadaba et al., 1989). Kinematic data were obtained with an 8-camera optical LED system (Impulse, Phase Space; San Leandro, CA) at a sampling rate of 240 Hz. Kinetic data were obtained using an embedded Bertec forceplate (Type 4060-130; Bertec Corporation, Columbus OH, USA) and were sampled at a rate of 1000 Hz. External joint moments were obtained using standard inverse dynamic equations and were normalized to body weight and height. Kinematic and kinetic instrumentation were interfaced with MotionMonitor software and were manually synced by a pulse trigger during each trial.

Data Handling and Processing. Motion capture began two seconds before initial ground contact, defined as the point at which the vertical ground reaction force exceeded 10N, and continued for three seconds after initial ground contact, for a total of five seconds. All flexions, adductions, and internal rotations were defined as positive angles. All extensions, abductions, and external rotations were defined as negative angles. To determine the appropriate filter for kinematic and kinetic data, a residual analysis was conducted on the ground reaction force (GRF) in a subset of the trials. Because the signal to noise ratio is dependent on the physical motion capture system and the velocities being captured and should therefore be stable across participants, using a subset of the data is more than adequate to determine the appropriate filtering frequency (E. Kristianslund et al., 2012; Eirik Kristianslund et al., 2012; Winter, 1990). To conduct the residual analysis, raw trials from four randomly selected participants were used. Each trial was filtered in MATLAB (MathWorks, Inc., Natwick, MA) under a series of low-pass filters, yielding multiple "versions" of each GRF time series. A sum of squares was computed for each non-raw version at each time point. This can be represented as $(raw_X\text{-}filtered_{yx})^2$, where *x* is the frame of data and *y* is the filtering frequency. Once the sums of squares were computed for each frequency, residuals were obtained. For each filtering frequency, the residual is defined as the square root of the mean sum of squares across all time points (Winter, 1990). Finally, to determine the proper filtering frequency, separate plots were generated for each GRF residual. Visual inspection of the plots indicated that the optimum ratio of signal distortion to noise was approximately at 10 Hz. Therefore, all kinetic and kinematic data were filtered with a 10 Hz low-pass, zero-lag, 2nd order Butterworth filter. Inverse dynamics were then computed to determine external moments for each joint (Gagnon & Gagnon, 1992), which were then normalized to each participant's height (meters) and mass (kilograms). Kinematic and kinetic data were then exported to MATLAB (MathWorks, Inc., Natwick, MA) for further reduction using custom-written script.

For reduction of MVIC and EMG data, the peak force (N) for each MVIC condition was multiplied by the moment arm length (as determined by Dempster's data and accounting for placement of the dynamometer) (Dempster, 1955), then divided by participant body mass, resulting in a normalized torque, $N \cdot m \cdot kg^{-1}$. These torque values were used to represent maximum torque generation. To obtain MVIC muscle activation amplitude, the EMG data were filtered in MATLAB using a band-pass 20-350 Hz, 4th order, zero-lag, Butterworth filter with full-wave rectification, and were processed using a root mean squared (RMS) algorithm with a 25-millisecond time constant. The peak RMS EMG amplitude was extracted from each trial and averaged to obtain the mean peak RMS amplitude for hip abduction (GMed) and hip extension (GMax) that was used to normalize the EMG signal (% MVIC) obtained during the landing task.

To prepare biomechanical data for SPM analysis, the landing phase (initial ground contact until the point of maximum knee flexion) was extracted for all biomechanical data. Custom MATLAB code was written to interpolate the values for each biomechanical variable during landing phase using a time series curve of 101 equally spaced data points.

Statistical Approach. To address our hypotheses, statistical parametric mapping (SPM) canonical correlation analyses (CCA) were used. SPM is a recently developed class of analyses whereby group differences and correlations can be examined across a time vector. The CCA in particular, is a multivariate SPM analysis. One benefit of canonical correlation analyses is that it can analyze multiple dependent variables at the same time (i.e. the linear combination of the components of functional valgus collapse; knee abduction, knee internal rotation, hip adduction, hip internal rotation). However, a drawback to this procedure is that it can only analyze one independent variable at a time. Therefore, separate SPM CCAs were conducted to determine the influence of *each* independent variable on a composite curve (linear combination) of functional knee valgus. Analyses were separated by sex and kinetics vs kinematics. In this way, the time points at which each independent variable influences the overall patterns of functional valgus collapse can be identified. Descriptive graphs were created and included (e.g. Figure 6.3) for completeness, but were not directly used in the interpretation of SPM CCAs. To properly interpret results from an SPM CCA analysis, SPM inferential graphs supplemented with GLM complementary analyses were necessary. Examples of inferential graphs are presented in Figures 6.4a-g. For each set of inferential graphs, independent variables are depicted separately. The dotted line in each graph represents the critical X^2-statistic. When the observed value, as represented by the SPM X^2 statistic, exceeds the dotted line, it creates a supra-threshold cluster, above which lies statistical significance. Thus, a supra-threshold cluster spans the time frame during which statistical significance is observed. For example, in Figure 6.4a-g, all statistical significance was contained in four supra-threshold clusters (from 86-95% and at 100% in c, at 98% in f, and from 48-52% in g). Percent landing phase is along the x-axis of the inferential graphs. In this way, the *moment(s)* in time at which significant associations occur during the landing phase can be identified.

Complementary General Linear Model (GLM) canonical correlation analyses were then used to determine the directionality and strength of significant relationships within each significant time frame (supra-threshold cluster), as well as provide information regarding the loading pattern of combined anatomical and neuromuscular variables on each component of functional valgus collapse. The benefit of a GLM canonical correlation is that multiple predictors and multiple dependent variables can be used. The disadvantage is that a time vector cannot be included; analysis must be contained to a single data point. In this way, we could determine the extent to which gluteal function influenced each biomechanical component at pre-determined time points, while also controlling for the other suppressor variables. Therefore, GLM canonical correlation analysis was conducted within each supra-threshold cluster (area of statistical significance) identified by SPM CCA. For instance, for the four supra-threshold clusters in Figure 6.4a-g, four GLM canonical correlations were conducted (e.g. Table 6.2) at the time points corresponding to the peak of each supra-threshold cluster (e.g at 91%, 98% and 100% in Figure 6.4c). GLM canonical correlations were conducted in stepwise fashion to reflect the influence of sagittal plane hip and knee position and transverse plane femoral alignment on lower extremity biomechanics (step 1-control variables; step 2-anatomical variables; step 3-neuromuscular variables). To further understand the particular influences of neuromuscular variables on individual components of functional valgus collapse, univariate results of the omnibus GLM canonical correlations were also examined (e.g. Table 6.3). Collectively, this series of analyses allowed us to identify the relevant time points in the landing to examine, and then to parse out the particular associations between gluteal muscle function, femoral alignment, and each component of functional valgus collapse. All SPM analyses were conducted using MATLAB code developed by Todd Pataky (Pataky, T., 2016, www.spm1d.org).

Results

Descriptive statistics for anatomical and neuromuscular variables in females and males are presented in Table 6.1. As expected, females had greater femoral anteversion and ROM_{IR} and less hip extension and hip abduction peak torque, and less gluteus medius muscle activation, thus confirming the need for sex-stratified analysis. Results will first be presented for female kinematics and kinetics, followed by male kinematics and kinetics.

Table 6.1. Descriptive Statistics for Independent Variables in Males and Females.

	Fem. Ant. (°)§	ROM_{IR} (°)§	ROM_{ER} (°)	HEX_{PTQ} ($N{\cdot}m{\cdot}kg^{-1}$)*	$HABD_{PTQ}$ ($N{\cdot}m{\cdot}kg^{-1}$)*	$HEX_{\%MVIC}$ (%)	$HABD_{\%MVIC}$ (%)*
Females	9.4±4.5	30.4±10.3	47.8±7.6	.86±.21	1.15±.24	37.2±31.4	42.2±23.7
Males	3.4±4.4	20.1±8.5	49.2±6.4	1.04±.26	1.49±.29	31.9±29.2	80.6±114.6
Totals	6.4±4.5	25.3±9.4	48.5±7.0	.95±.24	1.32±.27	34.6±30.3	61.4±69.2

Fem. Ant. = femoral anteversion; ROM_{IR} = internal rotation ROM; ROM_{ER} = external rotation ROM; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; *M > F at p <.05; § F > M at p <.05

Female Kinematic Valgus Collapse. SPM revealed that gluteus medius activation predicted the 4-component valgus collapse linear combination from 48-52% of the landing phase ($X^2_{crit=12.45} = 12.45$; $p = .05$) (Figure 6.4g). After accounting for all variables, post-hoc omnibus and univariate analyses did not identify gluteus medius activation as a significant predictor. Instead they indicated that at the 50% (peak) time point, the most strongly predicted component was knee rotation ($R^2 = .37$, $p = .04$) (Tables 6.2 and 6.3), such that higher ROM_{IR} (standardized beta (β_{st}) = .83) and greater gluteus maximus activation ($\beta_{st} = .37$) were associated with greater knee internal rotation (Table 6.3).

Later in the landing ROM_{ER} predicted the 4-component valgus collapse ($X^2_{crit=12.47}$ =12.47) from 86-95% ($p = .05$), 97-98% ($p = .05$), and at 100% of the landing phase ($p = .05$)

(Figure 6.4c), while gluteus maximus activation predicated valgus collapse only at the 98% mark ($X^2_{crit=12.48}$ = 12.48; p = .05) (Figure 6.4f). GLM post-hoc analyses revealed that at 91%, knee abduction was the most strongly predicted component (R^2 = .37, p = .04), such that greater ROM_{ER} (β_{st} = .53) was associated with greater knee abduction. At 98% and at 100%, hip adduction was the most strongly predicted component (R^2 = .35, p = .05 and .07, respectively), although no significant individual predictors were identified (Table 6.3).

Figure 6.3. Descriptive Curves of *Kinematic* Functional Valgus Collapse Components (Frontal and Transverse Plane Hip and Knee Motion) in *Females* during a Single-Leg Forward Landing Task.

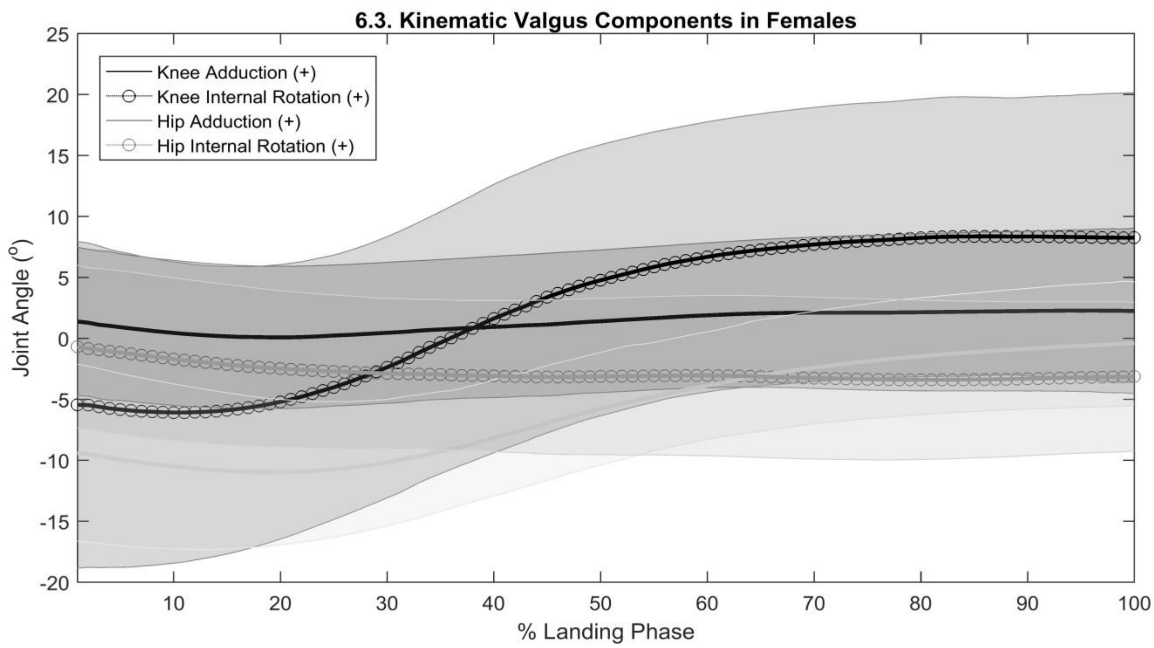


Figure 6.4. Inferential Results of SPM Canonical Correlation Analyses: the Relationship between Individual Predictors and a 4-Component *Kinematic* Valgus Collapse Combination in *Females*.

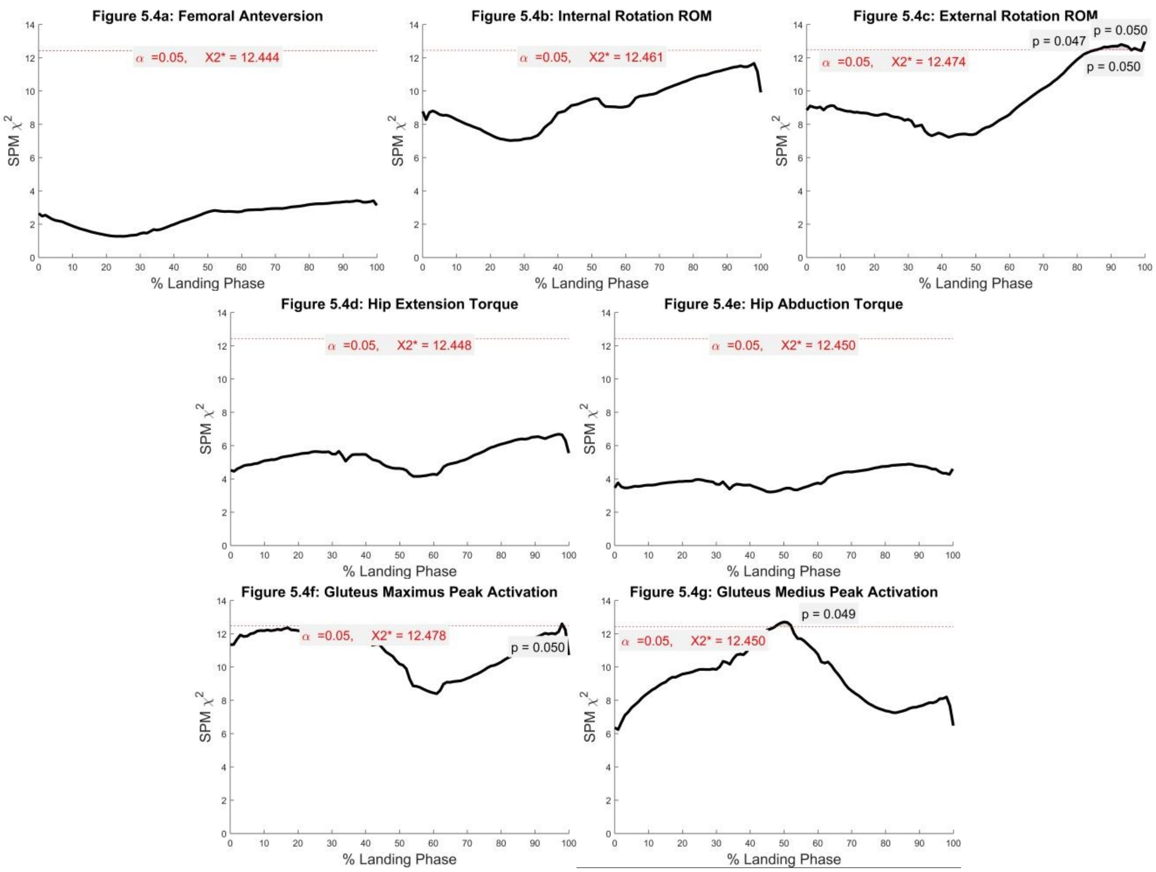


Table 6.2. Post-Hoc Canonical Correlation Omnibus Results Detailing the Contributions of Control Variables, Anatomical Variables, and Neuromuscular Variables to a 4-Component Kinematic Valgus Collapse Combination at Selected Time Points during the Landing Phase in Females.

			Standardized Canonical Coefficients											
			Control Variables		Anatomical Variables			Neuromuscular Variables				Functional Valgus Collapse Variables		
Time point	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Fem. Ant.	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC	Knee Abd	Knee Rot	Hip Add	Hip Rot
50%	1	.28(.01)	-.99	.24								-.67*	-.42	-.24	.55*
	2	.27(.01)	-.81	.06	-.02	.14	-.49					-.72*	-.37	-.32	.54*
	3	.25(.01)	-.70	.09	.24	-.55	-.59	.46	-.40	-.14	-.42	-.64	-.53*	-.45	.46
91%	1	.36(.19)	.76	-.68								.44	.23	-.45	-.53
	2	.31(.01)	.59	-.24	.16	-.37	.67					.76*	.26	.01	-.57*
	3	.31(.01)	.57	-.24	-.08	.39	.77	-.35	.51	.31	.14	.71*	.50	.03	-.50
98%	1	.35(.16)	.75	-.71								.37	.36	-.50*	-.46
	2	.30(.01)	.56	-.27	.18	-.38	.67					.70*	.38	-.04	-.60*
	3	.29(.01)	-.58	.23	.09	-.39	-.74	.34	-.45	-.39	-.12	-.65	-.62	-.04*	.52
100%	1	.37(.34)	.70	-.80								.29	.30	-.55	-.48
	2	.32(.04)	-.52	.30	-.21	.44	-.69					-.70*	-.21	.05	.63
	3	.33(.02)	.54	-.31	.00	.25	.77	-.29	.52	.33	.08	.66	.42	-.07	-.54

Fem. Ant. = femoral anteversion; ROMIR = internal rotation range of motion; ROMER = external rotation range of motion; HEXPTQ = hip extension peak torque; HABDPTQ = hip abduction peak torque; HEX%MVIC = hip extension activation amplitude; HABD%MVIC = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-); *significant at *p* <.05.

Table 6.3. Univariate Follow-Up Analyses Detailing the Contributions of Individual Predictors to each Component of Kinematic Valgus Collapse at Selected Time Points during the Landing Phase in Females.

				Standardized Beta Weights								
				Control Variables		Anatomical Variables			Neuromuscular Variables			
Time Point	Variable	Step	R^2 (p)	Knee Flexion	Hip Flexion	Fem. Ant.	ROM_{IR}	ROM_{ER}	HEX_{PTQ}	$HABD_{PTQ}$	$HEX_{\%MVIC}$	$HABD_{\%MVIC}$
50%	Knee Abd	1	.17(.02)	.42*	-.04							
		2	.27(.02)	.44*	.06	-.16	.19	.32				
		3	.34(.08)	.41*	-.02	-.14	.35	.42*	-.31	.36	-.18	.10
	Knee Rot	1	.10(.10)	.31*	-.11							
		2	.20(.11)	.35*	-.02	-.15	.37	.29				
		3	.37(.04)	.33*	-.04	-.31	.83*	.32	-.17	.28	.37*	.21
	Hip Add	1	.00(.91)	-.03	.06							
		2	.06(.74)	-.06	.02	.04	-.30	-.11				
		3	.26(.25)	-.10	.07	-.15	-.04	-.01	-.24	-.12	-.19	.43
	Hip Rot	1	.15(.03)	-.37*	.16							
		2	.28(.02)	-.29*	.15	-.32	.51*	-.09				
		3	.31(.12)	-.31	.14	-.31	.48	-.04	-.10	.03	-.24	.10
91%	Knee Abd	1	.11(.09)	.29	-.16							
		2	.31(.01)	.36*	-.06	-.04	-.04	.43*				
		3	.37(.04)	.30	-.12	-.02	.15	.53*	-.29	.39	-.12	.10
	Knee Rot	1	.03(.54)	.16	-.05							
		2	.18(.16)	.24	.04	-.26	.49*	.34*				
		3	.34(.07)	.26	.01	-.39	.95*	.38*	-.25	.33	.41*	.09
	Hip Add	1	.13(.06)	-.26	.25							
		2	.18(.16)	-.28	.19	-.003	-.17	-.25				
		3	.35(.06)	-.26	.25	-.15	-.08	-.23	-.11	-.30	-.11	.23
	Hip Rot	1	.08(.17)	-.16	.24							
		2	.26(.03)	-.12	.23	-.36	.60*	-.08				
		3	.31(.12)	-.16	.20	-.31	.58	-.03	-.14	.12	-.23	.03
98%	Knee Abd	1	.08(.17)	.23	-.18							
		2	.25(.04)	.29	-.09	-.00	-.10	.37*				
		3	.31(.11)	.23	-.15	.02	.11	.48*	-.32	.40	-.13	.10
	Knee Rot	1	.06(.26)	.25	-.05							
		2	.21(.10)	.31*	.03	-.27	.48*	.33				

		3	.35(.06)	.35*	.01	-.39	.87*	.34	-.20	.26	.42*	.04
	Hip Add	1	.14(.04)	-.26	.29*							
		2	.20(.11)	-.28	.22	-.00	-.17	-.26				
		3	.35(.05)	-.25	.27	-.15	-.08	-.24	-.09	-.29	-.08	.22
	Hip Rot	1	.06(.26)	-.11	.23							
		2	.26(.03)	-.08	.21	-.38	.63*	-.10				
		3	.31(.11)	-.12	.18	-.33	.64	-.03	-.18	.17	-.21	.03
100%	Knee Abd	1	.06(.25)	.22	-.16							
		2	.26(.03)	.31*	-.04	-.06	-.00	.45*				
		3	.34(.07)	.26	-.11	-.01	.20	.55*	-.26	.44	-.10	.08
	Knee Rot	1	.04(.42)	.19	-.11							
		2	.13(.36)	.24	-.02	-.24	.38	.27				
		3	.33(.09)	.29	-.02	-.40	.85*	.30	-.29	.26	.44*	.09
	Hip Add	1	.12(.07)	-.26	.27							
		2	.17(.19)	-.28	.19	-.05	-.07	-.27				
		3	.35(.07)	-.29	.24	-.19	.01	-.26	-.11	-.30	-.09	.24
	Hip Rot	1	.07(.24)	-.09	.25							
		2	.25(.04)	-.09	.25	-.35	.58*	-.11				
		3	.29(.18)	-.12	.21	-.31	.58	-.06	-.16	.11	-.20	.02

Fem. Ant. = femoral anteversion; ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at p <.05.

Female Kinetic Valgus Collapse. SPM revealed that ROM_{ER} predicted the kinetic 4-component valgus collapse linear combination ($X^2_{crit=15.58} = 15.76$) from 7-8% (p =.05), at 18% (p =.05) and at the 20% mark (p =.05) (Figure 6.6c). Accounting for all variables, GLM post-hoc omnibus and univariate analyses (Table 6.4 and 6.5) indicated that at 8% and at 18%, anatomical and muscle characteristics were most strongly associated with external hip adduction moment (R^2 = .33, p = .08; R^2 = .34, p = .06, respectively). At 8%, greater hip adduction moment was associated with greater ROM_{ER} (β_{st} = .53), while at 18%, greater hip adduction moment was most associated with greater gluteus medius activation (β_{st} = .50) (Table 6.5).

Figure 6.5. Descriptive Curves of *Kinetic* Functional Valgus Collapse Components (Frontal and Transverse Plane Hip and Knee Moments) in *Females* during a Single-Leg Forward Landing Task.

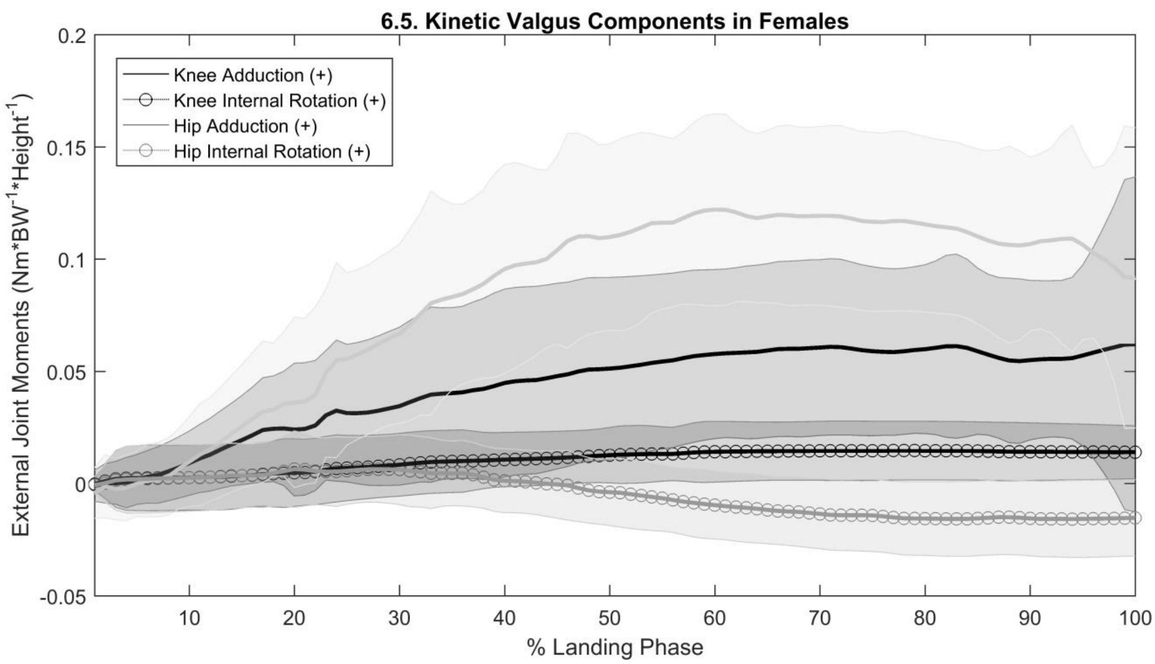


Figure 6.6. Inferential Results of SPM Canonical Correlation Analyses: the Relationship between Individual Predictors and a 4-Component *Kinetic* Valgus Collapse Combination in *Females*.

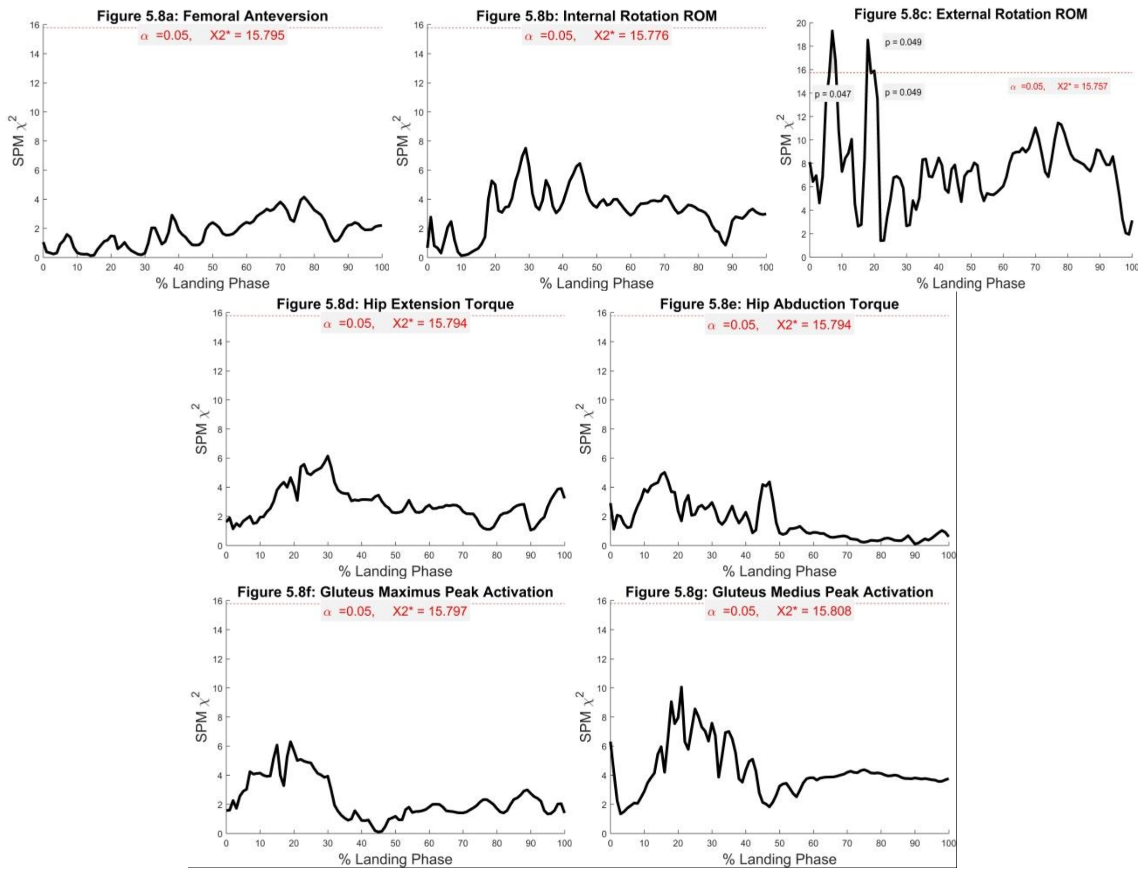


Table 6.4. Post-Hoc Canonical Correlation Omnibus Results Detailing the Contributions of Control Variables, Anatomical Variables, and Neuromuscular Variables to a 4-Component Kinetic Valgus Collapse Combination at Selected Time Points during the Landing Phase in Females.

			Standardized Canonical Coefficients												
			Control Variables		Anatomical Variables			Neuromuscular Variables				Functional Valgus Collapse Variables			
Time point	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Fem. Ant.	ROM $_{IR}$	ROM $_{ER}$	HEX $_{PTQ}$	HABD $_{PTQ}$	HEX $_{\%MVIC}$	HABD $_{\%MVIC}$	Knee Abd	Knee Rot	Hip Add	Hip Rot
8%	1	.13(.07)	-.52	.84								.80	-3.72	.20	4.10
	2	.16(.03)	-.19	-.41	-.14	-.13	-1.07					.82	-2.64	-1.04*	3.42
	3	.14(.14)	-.12	-.33	.03	-.70	-1.14	.62	-.38	-.06	-.30	.89	-2.71	-1.03	3.52
18%	1	.10(.01)	-.60	.78								.68	-1.98	.28*	1.92
	2	.06(.003)	-.42	.02	-.55	.46	-.79					.78	-1.82	-.58	2.33
	3	.07(.01)	-.33	-.05	-.39	.03	-.94	.52	-.24	.25	-.43	.77	-1.69	-.72	2.24

Fem. Ant. = femoral anteversion; ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at p <.05.

Table 6.5. Univariate Follow-Up Analyses Detailing the Contributions of Individual Predictors to each Component of Kinetic Valgus Collapse at Selected Time Points during the Landing Phase in Females.

				Standardized Beta Weights								
				Control Variables		Anatomical Variables			Neuromuscular Variables			
Time Point	Variable	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Femoral Anteversion	ROM_{IR}	ROM_{ER}	HEX_{PTQ}	$HABD_{PTQ}$	$HEX_{\%MVIC}$	$HABD_{\%MVIC}$
8%	Knee Abd	1	.04(.45)	-.07	.18							
		2	.07(.71)	-.11	.21	-.10	-.02	.11				
		3	.15(.71)	-.14	.11	-.05	.04	.20	-.26	.28	-.26	.04
	Knee Rot	1	.01(.88)	-.03	-.07							
		2	.04(.90)	.01	-.13	-.03	-.01	-.21				
		3	.10(.92)	.02	-.12	-.10	-.02	-.22	.20	-.08	-.07	.19
	Hip Add	1	.12(.06)	.01	.35*							
		2	.30(.01)	-.09	.48*	-.07	.09	.46*				
		3	.33(.08)	-.12	.49*	-.17	.30	.53*	-.21	.08	-.06	.19
	Hip Rot	1	.01(.88)	-.08	-.02							
		2	.05(.82)	-.03	-.08	-.05	-.00	-.25				
		3	.11(.87)	-.01	-.04	-.10	-.08	-.30	.29	-.19	-.01	.13
18%	Knee Abd	1	.07(.23)	.02	.26							
		2	.09(.59)	.00	.31	.03	.02	.16				
		3	.19(.53)	-.07	.19	.00	.17	.27	-.16	.33	-.31	.27
	Knee Rot	1	.01(.73)	.12	-.03							
		2	.04(.89)	.14	-.08	.02	-.06	-.18				
		3	.11(.88)	.10	-.07	-.05	-.06	-.18	.21	-.08	-.09	.22
	Hip Add	1	.13(.05)	.05	.36*							
		2	.20(.10)	.02	.42*	.32	-.24	.22				
		3	.34(.06)	-.06	.41*	.13	.08	.31	-.06	.13	-.15	.50*
	Hip Rot	1	.01(.84)	-.07	.06							
		2	.13(.36)	-.02	-.05	-.08	.03	-.38*				
		3	.21(.43)	-.03	-.00	-.13	-.07	-.45*	.37	-.22	.08	.09

ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at p <.05.

Male Kinematic Valgus Collapse. SPM revealed that the 4-component valgus collapse linear combination in males was associated with ROM_{IR} at 0% time point (p = .05) and from 2-3% (p = .05) of the landing phase ($X^2_{crit=12.19}$ = 12.19) (Figure 6.8b); hip abduction torque from 15-52% of the landing phase ($X^2_{crit=12.20}$ = 12.20; p = .03) (Figure 6.8e); and femoral anteversion from 69-100% of the landing phase ($X^2_{crit=12.19}$ = 12.19; p = .04) (Figure 6.8a). Post-hoc omnibus and univariate analyses (Tables 6.6 and 6.7) indicated that at 0% and from 2-3%, greater ROM_{IR} (β_{st} = -.55) and greater ROM_{ER} (β_{st} = -.33) contributed to greater knee abduction (R^2 = .32, p = .10), while greater ROM_{ER} (β_{st} = .56) predicted greater hip adduction at 0% and 3% (R^2 = .34 and .35, p = .07 and .06, respectively). From 15-52% (using the 40% time point) of the landing phase, less peak hip abduction torque (β_{st} = -.47) was associated with greater knee internal rotation (R^2 = .39, p = .03). Lastly, from 69-100% (using 100% time point), less femoral anteversion (β_{st} = -.43) and less peak hip abduction torque (β_{st} = -.49) were associated with greater knee internal rotation (R^2 = .40, p = .02).

Figure 6.7. Descriptive Curves of *Kinematic* Functional Valgus Collapse Components (Frontal and Transverse Plane Hip and Knee Motion) in *Males* during a Single-Leg Forward Landing Task.

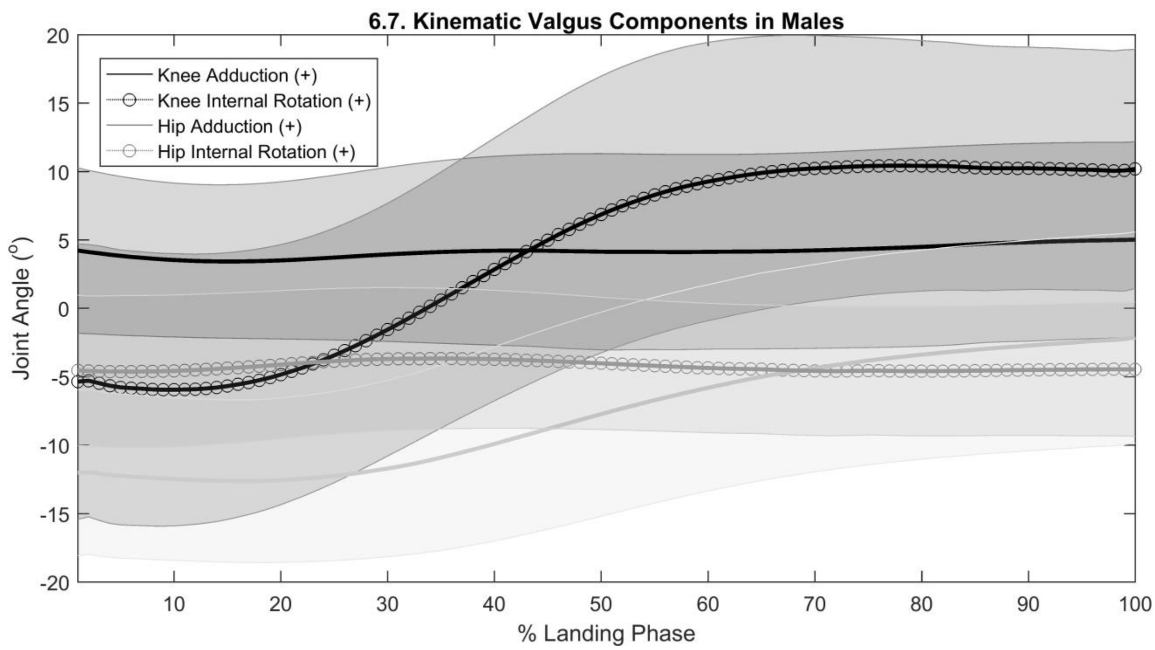


Figure 6.8. Inferential Results of SPM Canonical Correlation Analyses: the Relationship between Individual Predictors and a 4-Component *Kinematic* Valgus Collapse Combination in *Males*.

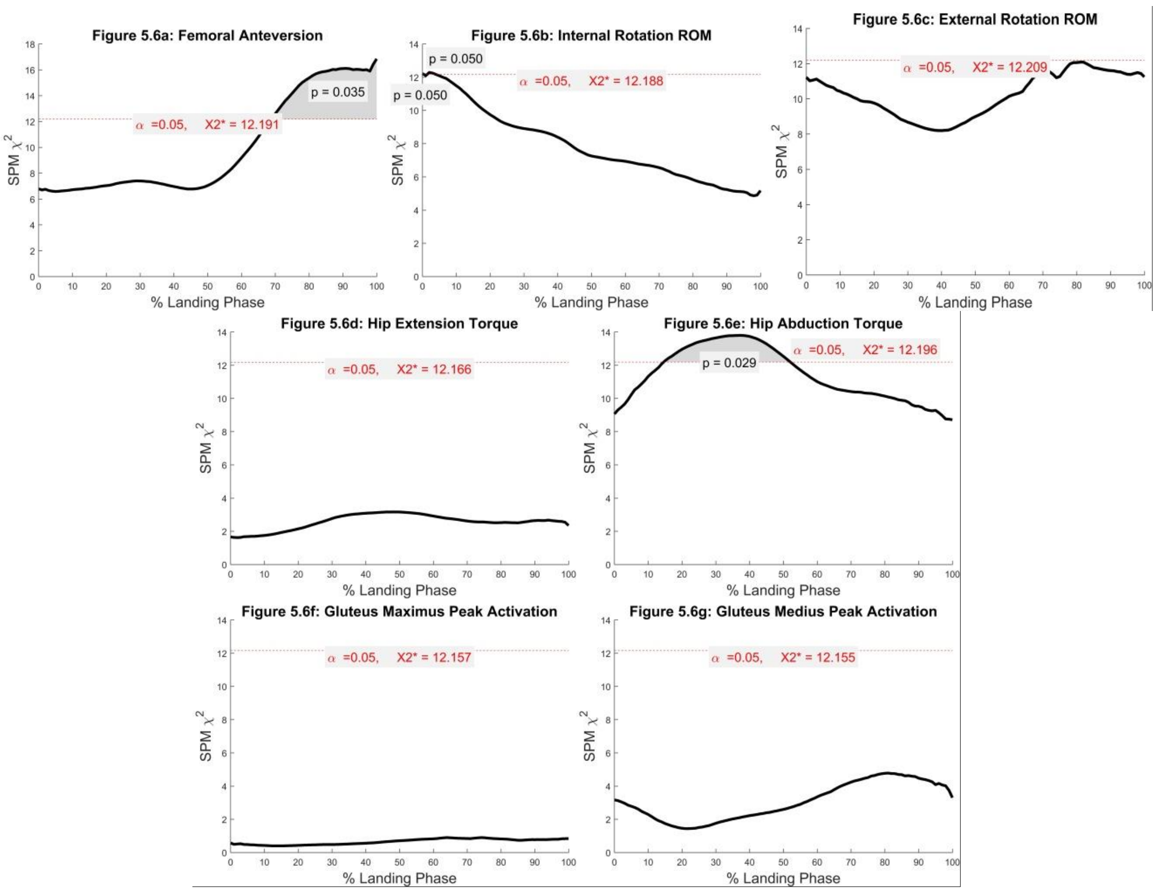


Table 6.6. Post-Hoc Canonical Correlation Omnibus Results Detailing the Contributions of Control Variables, Anatomical Variables, and Neuromuscular Variables to a 4-Component Kinematic Valgus Collapse Combination at Selected Time Points during the Landing Phase in Males.

			Standardized Canonical Coefficients												
			Control Variables		Anatomical Variables			Neuromuscular Variables				Functional Valgus Collapse Variables			
Time point	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Fem. Ant.	ROM_{IR}	ROM_{ER}	HEX_{PTQ}	$HABD_{PTQ}$	$HEX_{\%MVIC}$	$HABD_{\%MVIC}$	Knee Abd	Knee Rot	Hip Add	Hip Rot
0%	1	.13(.06)	.82	.40								.85	.68	.68	.61
	2	.11(<.001)	.01	-.62	-.36	.72	-.37					-1.06*	.19	-1.08*	.11
	3	.17(.01)	-.27	.44	.33	-.18	.70	-.40	.51	.11	.14	.45	-.67	.96	-.45
3%	1	.13(.06)	.86	.36								.80	.70	.65	.66
	2	.11(<.001)	.02	-.60	-.40	.73	-.36					-1.01*	.25	-1.07*	.13
	3	.17(.01)	-.27	.41	.34	-.15	.69	-.41	.55	.12	.13	.36	-.71	.92	-.47
40%	1	.22(.34)	-1.04	.16								-.26	-.90	-.59	-.42
	2	.28(.01)	.42	-.18	-.41	.92	.55					-.85*	.67	-.17	-.14
	3	.19(.02)	.08	-.02	-.41	.13	-.34	.52	-.86	-.44	.29	-.34	.94*	-.50	.46
100%	1	.26(.80)	-.95	-.21								-.29	-.82	-.75	.37
	2	.32(.01)	-.33	-.14	.78	-.47	-.50					.57	-.41	.24*	.61*
	3	.30(.03)	.16	.18	-.82	.12	-.03	.24	-.59	-.26	.14	-.51	.78*	-.43	-.15

ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at p <.05.

Table 6.7. Univariate Follow-Up Analyses Detailing the Contributions of Individual Predictors to each Component of Kinematic Valgus Collapse at Selected Time Points during the Landing Phase in Males.

				Standardized Beta Weights								
				Control Variables		Anatomical Variables			Neuromuscular Variables			
Time Point	Variable	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Femoral Anteversion	ROM IR	ROM ER	HEX PTQ	HABD PTQ	HEX %MVIC	HABD %MVIC
0%	Knee Abd	1	.04(.42)	.19	.04							
		2	.31(.01)	.17	.10	-.04	-.53*	-.31				
		3	.32(.10)	.17	.10	-.04	-.55*	-.33*	.11	-.06	-.08	.05
	Knee Rot	1	.07(.21)	.28	-.05							
		2	.22(.08)	.25	-.02	-.32*	.25	.18				
		3	.30(.13)	.16	.02	-.29	.11	.10	.07	-.33	-.14	-.00
	Hip Add	1	.06(.26)	-.05	.26							
		2	.29(.02)	-.11	.26	.21	.14	.52*				
		3	.34(.07)	-.07	.28	.14	.17	.56*	-.22	.15	.00	.07
	Hip Rot	1	.02(.62)	.14	.03							
		2	.17(.19)	.21	-.02	.13	-.04	-.34*				
		3	.19(.51)	.20	-.03	.17	-.05	-.37*	.16	-.07	-.04	-.03
3%	Knee Abd	1	.03(.48)	.18	.03							
		2	.31(.01)	.16	.09	-.05	-.53*	-.32*				
		3	.32(.10)	.16	.09	-.05	-.55*	-.33*	.10	-.05	-.07	.05
	Knee Rot	1	.07(.22)	.27	-.05							
		2	.21(.08)	.24	-.02	-.31	.26	.19				
		3	.30(.13)	.16	.01	-.28	.12	.11	.07	-.34	-.13	.00
	Hip Add	1	.06(.25)	-.02	.26							
		2	.29(.02)	-.08	.26	.22	.13	.51*				
		3	.35(.06)	-.04	.28	.15	.17	.56*	-.22	.15	.00	.07
	Hip Rot	1	.03(.56)	.16	.02							
		2	.18(.17)	.23	-.02	.12	-.04	-.35*				
		3	.20(.48)	.22	-.03	.15	-.04	-.38*	.16	-.06	-.04	-.03
40%	Knee Abd	1	.02(.61)	-.08	.16							
		2	.24(.05)	-.08	.17	-.00	-.48*	-.33*				
		3	.28(.20)	-.15	.16	.00	-.53*	-.35*	.17	-.16	-.08	.11
	Knee Rot	1	.12(.07)	.34*	-.01							
		2	.24(.05)	.35*	-.00	-.28	.29	.12				
		3	.39(.03)	.16	.04	-.28	.11	.03	.24	-.47*	-.27	.23
	Hip Add	1	.04(.47)	.20	-.08							

		2	.22(.08)	.18	-.10	.28	.20	.39*				
		3	.31(.12)	.28	-.10	.21	.27	.46*	-.30	.24	.08	.00
	Hip Rot	1	.02(.72)	.08	-.13							
		2	.16(.23)	.08	-.13	.14	-.03	-.32				
		3	.20(.49)	-.02	-.11	.16	-.12	-.37*	.13	-.24	-.11	.07
100%	Knee Abd	1	.00(.92)	-.07	.01							
		2	.18(.15)	-.07	.01	.05	-.38*	-.35*				
		3	.21(.41)	-.11	-.01	.09	-.43*	-.38*	.17	-.14	.03	.02
	Knee Rot	1	.05(.32)	.23	-.01							
		2	.22(.07)	.24	.07	-.44*	.21	-.00				
		3	.40(.02)	.09	.13	-.43*	.00	-.09	.27	-.49*	-.25	.20
	Hip Add	1	.01(.75)	.05	.10							
		2	.25(.04)	.06	.01	.33*	.27	.40*				
		3	.32(.10)	.12	.00	.24	.37*	.47*	-.15	.25	-.06	.12
	Hip Rot	1	.04(.45)	-.19	.01							
		2	.31(.01)	-.14	-.12	.35*	-.10	-.36*				
		3	.35(.05)	-.20	-.08	.38*	-.14	-.40*	.20	-.13	-.11	-.01

ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at *p* <.05.

Male Kinetic Valgus Collapse. SPM analysis revealed that ROM_{IR} was associated with the kinetic 4-component linear combination ($X^2_{crit=14.87} = 14.87$) from 3-9% (p =.04) of the landing phase (Figure 6.10b). Post-hoc omnibus and univariate analyses (Tables 6.8 and 6.9) indicated that at 6%, both knee abduction moment ($R^2 = .42$, $p = .01$) and hip adduction moment ($R^2 = .53$, $p = .001$) were significantly predicted. Specifically, greater ROM_{IR} ($\beta_{st} = -.52$) and greater hip extension peak torque ($\beta_{st} = -.37$) were associated with greater knee abduction moment, while greater ROM_{IR} ($\beta_{st} = -.33$), lesser ROM_{ER} ($\beta_{st} = .46$), and greater hip extension peak torque ($\beta_{st} = -.46$) were associated with less hip adduction moment.

Figure 6.9. Descriptive Curves of *Kinetic* Functional Valgus Collapse Components (Frontal and Transverse Plane Hip and Knee Moments) in *Males* during a Single-Leg Forward Landing Task.

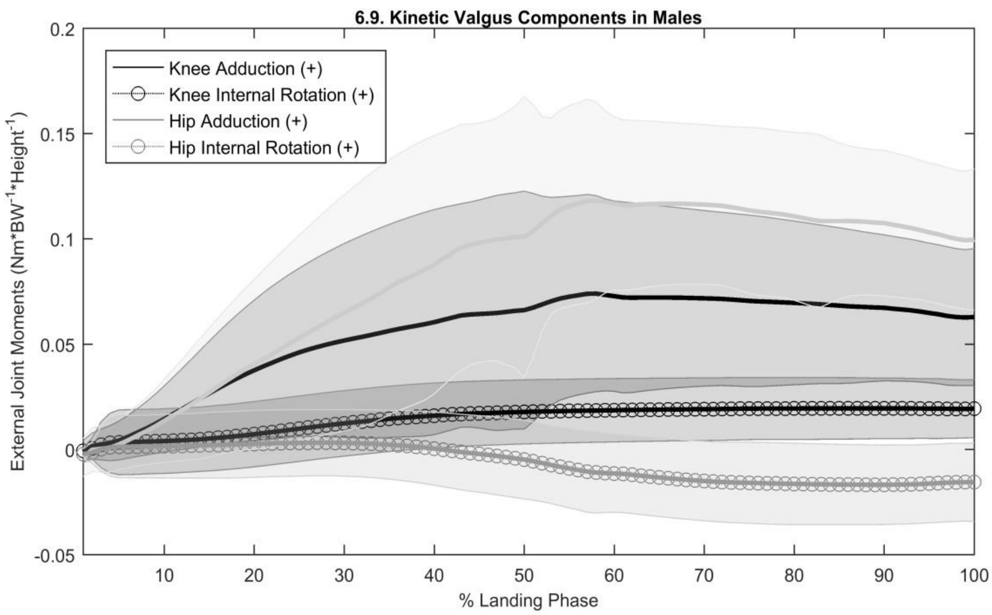


Figure 6.10. Inferential Results of SPM Canonical Correlation Analyses: the Relationship between Individual Predictors and a 4-Component *Kinetic* Valgus Collapse Combination in *Males*.

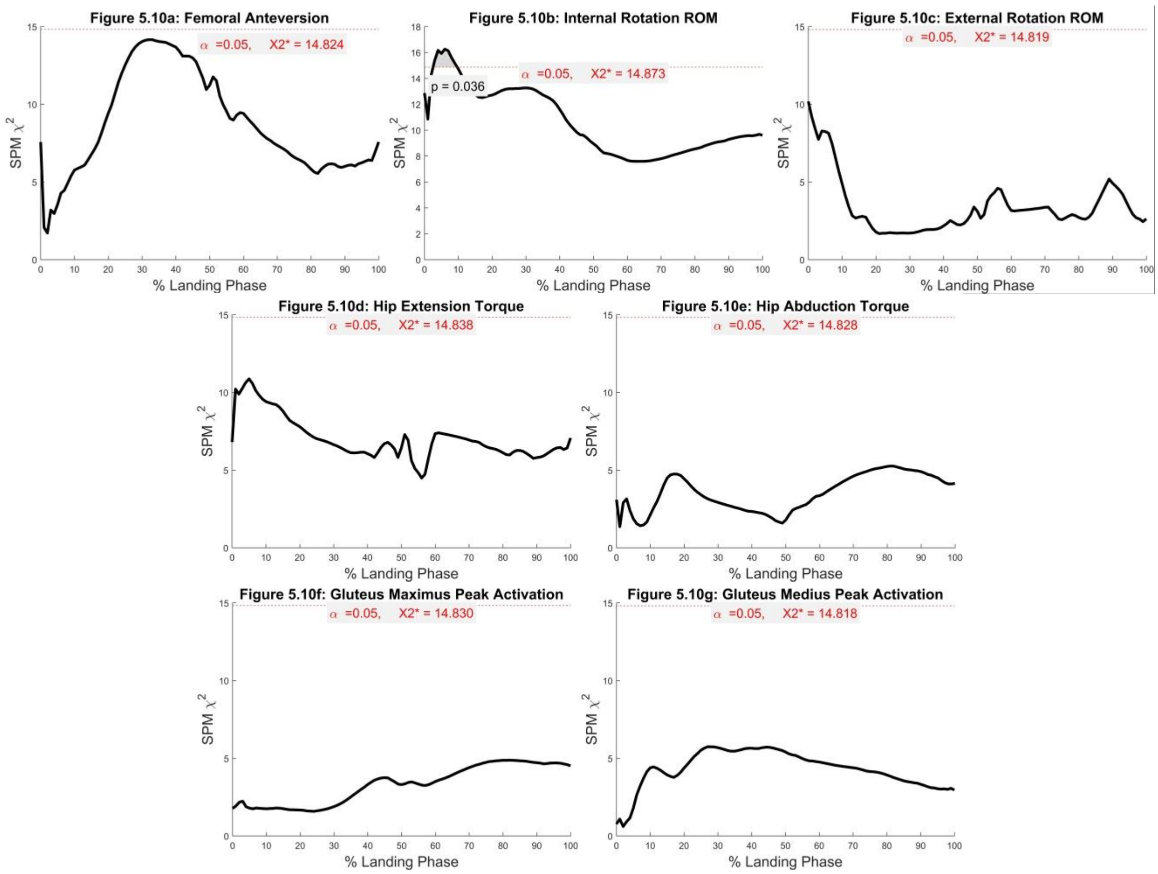


Table 6.8. Post-Hoc Canonical Correlation Omnibus Results Detailing the Contributions of Control Variables, Anatomical Variables, and Neuromuscular Variables to a 4-Component Kinetic Valgus Collapse Combination at Selected Time Points during the Landing Phase in Males.

			Standardized Canonical Coefficients												
			Control Variables		Anatomical Variables			Neuromuscular Variables				Functional Valgus Collapse Variables			
Time point	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Femoral Anteversion	ROM_{IR}	ROM_{ER}	HEX_{PTQ}	$HABD_{PTQ}$	$HEX_{\%MVIC}$	$HABD_{\%MVIC}$	Knee Abd	Knee Rot	Hip Add	Hip Rot
6%	1	.05(.12)	-1.01	.16								1.34	-3.15	-.91	3.97
	2	.19(.002)	.43	-.12	-.36	-.76	-.62					.06*	4.27	-.23*	-4.84
	3	.42(.001)	.07	.14	.25	-.46	.52	-.72	.12	.32	.04	-.29*	-.68	1.08*	.12

ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at p <.05.

Table 6.9. Univariate Follow-Up Analyses Detailing the Contributions of Individual Predictors to each Component of Kinetic Valgus Collapse at Selected Time Points during the Landing Phase in Males.

				Standardized Beta Weights								
				Control Variables		Anatomical Variables			Neuromuscular Variables			
Time Point	Variable	Step	R^2 (*p*)	Knee Flexion	Hip Flexion	Femoral Anteversion	ROM $_{IR}$	ROM $_{ER}$	HEX $_{PTQ}$	HABD $_{PTQ}$	HEX $_{\%MVIC}$	HABD $_{\%MVIC}$
6%	Knee Abd	1	.02(.68)	-.13	.06							
		2	.25(.04)	-.21	.13	.10	-.45*	.14				
		3	.42(.01)	-.21	.14	.03	-.52*	.17	-.37*	-.03	.08	.12
	Knee Rot	1	.03(.55)	-.09	.16							
		2	.14(.32)	-.07	.12	-.01	.36*	.13				
		3	.27(.21)	-.11	.14	.04	.32	.06	.35*	-.18	-.19	-.03
	Hip Add	1	.03(.50)	.04	.17							
		2	.30(.01)	-.05	.21	.27	-.27	.42*				
		3	.54(.001)	-.06	.21	.21	-.33*	.46*	-.46*	-.01	.16	.05
	Hip Rot	1	.04(.45)	-.14	.16							
		2	.19(.12)	-.12	.11	-.02	.42*	.18				
		3	.32(.10)	-.16	.13	.08	.39*	.11	.34*	-.17	-.18	-.02

ROM_{IR} = ROM_{IR} = internal rotation range of motion; ROM_{ER} = external rotation range of motion; HEX_{PTQ} = hip extension peak torque; $HABD_{PTQ}$ = hip abduction peak torque; $HEX_{\%MVIC}$ = hip extension activation amplitude; $HABD_{\%MVIC}$ = hip abduction activation amplitude; kinematic flexions, adductions, and internal rotation are (+); kinematic extensions, abductions, and external rotations are (-);*significant at p <.05.

Discussion

The objective of this study was to explore the contributions of gluteal function a combined hip and knee motions contributing to functional valgus collapse over time using a multivariate approach. This approach offers a more integrated and holistic biomechanical approaches instead of isolating and independently evaluating individual joints and planes of motion. Specifically, we hypothesized that once controlling for sagittal plane hip and knee position and hip anatomy, individuals with greater functional valgus collapse would display less gluteal strength together with greater gluteal muscle activation in an effort to stabilize the lower extremity upon landing on a single leg. Our results only partially support these hypotheses.

Though we did observe significant relationships early in the landing phase in both sexes, these relationships were intermittent and were not observed for any sustained length of time. Nevertheless, these findings serve as a preliminary probe into the complexity of lower extremity biomechanics and how the neuromuscular system works with anatomy to influence overall movement patterns. Most importantly, this work has provided baseline information so that future researchers can conduct more explicit hypothesis-driven, *a priori* experiments.

Functional Valgus Collapse in Females. In our female cohort, the only significant associations we observed in the first half of the landing phase occurred from 7 - 8%, 18%, and 20% of the way through the landing phase. The average time of completion for females in the single-leg forward landing was 203.2 ± 33.5ms, so the observed relationships occurred from approximately 14.2 ± 2.3 - 16.3 ± 2.7, 36.6 ± 6.0, and 40.6 ± 6.7ms after initial ground contact, and may therefore be relevant to the 30-50ms after initial ground contact in which ACL injuries are reported to occur (Carlson et al., 2016; T Krosshaug et al., 2007; Tron Krosshaug et al., 2007). In total, these results revealed that early in the landing phase anatomical influences predominated, followed by gluteal muscle influences. To aid in interpretation, these findings are discussed in order of appearance during the landing.

At the 7 - 8% time point in the landing phase, sagittal plane hip and knee position and femoral alignment accounted for 16% of the total variability in kinetic functional valgus collapse (Table 6.4). Within the single component of greater hip adduction moment, 30% of the variance was explained by greater ROM_{ER} and greater hip flexion angle (Table 6.5). Though the addition of neuromuscular variables at this time point in the landing did not improve the model, it is possible that anatomical associations at the 7 – 8% time point influenced gluteal associations later in the landing phase. In addition to ROM_{ER}, our data indicated that greater hip flexion angle was also associated with greater hip adduction moment. Interestingly, as the hip moves into flexion, the internal rotation moment arm of the gluteus medius lengthens (Delp et al., 1999). Once this happens, the gluteus medius will not only act in the frontal plane (hence the observed greater external hip adduction moment), but will also work to control excessive ROM_{ER} via its lengthened internal rotation moment arm. Therefore, it is possible that females with excessive ROM_{ER} may display greater hip flexion in order to use the gluteus medius to bring the hip closer to a neutral transverse plane alignment. If this be the case, one would expect gluteus medius activation to exhibit similar effects on movement concurrently or quickly thereafter, which was the case.

At the 18 and 20% time points, although SPM identified ROM_{ER} as a predictor of hip adduction moment, the post-hoc canonical correlation revealed that greater gluteus medius activation and greater hip flexion angle were the primary predictors of greater hip adduction moment (R^2 = .34). The association between heightened gluteus medius activity and greater hip adduction moment agrees with previously established thought (Homan et al., 2013a; A.-D. Nguyen et al., 2011), and suggests that females ramp up their gluteus medius activity in an unsuccessful effort to overcome an external hip adduction moment. This idea is reinforced by Figure 6.5, which shows hip adduction moment rising rapidly from 8% - 18% of the landing phase. Though our data indicated that greater gluteus medius activity didn't necessarily correspond to a weaker muscle, it is possible that greater muscle activation is indicative of insufficiency in another aspect of function. It is also plausible that gluteus medius activity could be heightened as a result of excessive ROM_{ER} and increased hip flexion, culminating in a greater hip adduction moment.

Functional Valgus Collapse in Males. In our male cohort, multiple frontal plane relationships were observed early in the landing phase. The time of task completion for males was 207.9 ± 35.7ms. The observed associations occurred at initial ground contact and at approximately 4.2 ± .7 - 6.2 ± 1.1ms (2 - 3% time point) and 6.2 ± 1.1 - 18.7 ± 3.2ms (3 - 9% time point) after initial ground contact, and therefore could be relevant to the occurrence of ACL injury. At the hip, greater ROM_{ER} predicted greater hip adduction angle at 0% and from 2 - 3%, while from 3 – 9% of the landing phase, greater ROM_{IR}, less ROM_{ER}, and greater hip extension torque combined to predict smaller hip adduction moment. At the knee, greater ROM_{IR} and greater ROM_{ER} predicted greater knee abduction angle at 0% and from 2 - 3%. From 3 – 9%, both greater ROM_{IR} and greater peak hip extension torque predicted greater knee abduction moment.

The data obtained from our male cohort suggested that males with greater amounts of transverse plane hip ROM (greater ROM_{IR} and greater ROM_{ER}) made initial ground contact with riskier frontal plane kinematics, as evidenced by greater knee abduction and greater hip adduction at 0% and from 2 -3%. Though the authors are unaware of previous research to corroborate these findings in a male cohort, similar findings have been published in all female or mixed-sex cohorts

(D. R. Bell et al., 2008; A Nguyen et al., 2009), and suggests that an excess of available ROM may present a challenge for proper joint positioning. Because these relationships were observed at initial ground contact (0% and 2 - 3% of the landing phase), it may also suggest that excessive ROM compromised landing preparation. If so, this could represent a potential area for exploration, as flight phase mechanics may be readily modifiable. Further research is needed to determine the extent to which flight phase mechanics are influenced by postural characteristics.

Shortly after initial ground contact, from 3 – 9% of the landing phase (6.2 ± 1.1 – 18.7 ± 3.2ms), greater hip extension peak torque, in addition to greater ROM_{IR}, were associated with higher knee abduction moment and lower hip adduction moment. This suggested that in the presence of excess ROM, participants with stronger hip extensors attempted to exert forces about the joint, possibly in an attempt to provide greater joint stability. This strategy appeared to have been effective at the hip, as evidenced by the prediction of a smaller hip adduction moment. At the knee, however, abduction moment continued to increase, demonstrating that gross hip extension strength was not entirely effective for correcting frontal plane alignment once the effect of increased hip ROM had been established.

Of note, similar patterns were observed in both sexes, in that neuromuscular effects occurred just a few milliseconds after effects associated with femoral alignment. Further research is needed to verify the relative timing of anatomical and neuromuscular effects on biomechanics, but based on these preliminary findings, neuromuscular timing could represent an avenue for mitigation of the influence of suboptimal femoral alignment.

Comparison of Statistical Parametric Mapping and General Linear Model Canonical Correlation Analyses. There were appreciable differences in the results obtained using SPM and post-hoc General Linear Model (GLM) canonical correlation analyses. In particular, the SPM CCA indicated little involvement of gluteal function in functional valgus collapse, whereas the post-hoc analysis uncovered multiple significant associations that implicated gluteal function. This discrepancy is likely explained by the ability of traditional canonical correlation analyses to account for multiple independent variables. Once accounting for sagittal plane position and the influence of one's femoral alignment on functional valgus collapse, the role of gluteal function was clearer in our post-hoc analyses. This was in contrast with the SPM analysis, in which a single predictor was associated with a linear combination of four variables across time. This also highlighted the importance of accounting for sagittal plane position and femoral alignment when analyzing the role of the gluteal muscles in lower extremity movement. Without these control variables, the role of gluteal function was ambiguous. Though the temporal component may be an important variable to consider, the current SPM CCA is not powerful enough to adequately handle all relevant variables. At this point therefore, it may be preferable to sacrifice the temporal component and retain the ability to include multiple predictors than to only analyze single predictors over time. Nevertheless, given that SPM CCA is a developing analytical tool, it may yet grow into this capability.

Limitations. There were considerable limitations with this study. Namely, the inability to include multiple predictors in an SPM CCA limits the confidence with which we can attribute variance to specific predictors. This discrepancy was apparent in the differing results obtained between the SPM and GLM canonical correlations. Even so, SPM CCA was useful for identifying time points of interest, which could then be analyzed in greater detail using GLM canonical correlations. As such, this study still serves as an appropriate preliminary step in moving toward more integrated assessment of global movement patterns. A second limitation with this study was its power. As this study was originally powered to detect an R^2 increase in individual biomechanical variables with the addition of neuromuscular variables, it was not powered to include the addition of multiple dependent variables and time vector. While power analyses for multivariate SPM models aren't available, power to detect significant relationships associated with neuromuscular variables in the GLM univariate correlations ranged from .12 - .71 in females. In the male cohort, it ranged from .07 - .64, save for a single relationship which achieved a power of .82.

Conclusion

Despite the exploratory nature of this study, these findings provide rationale for the further use and development of integrated analyses which take into account not only the dependent nature of biomechanical variables, but also the temporal aspect of movement patterns. In both the male and female cohorts, femoral alignment displayed effects in the first few milliseconds of the landing phase. Neuromuscular factors began to display significant roles 10 – 20ms later, but were largely ineffective in correcting the already established movement strategies. These findings provide a baseline from which future work can implement more hypothesis-driven, *a priori* experiments. Lastly, while complementary analyses confirmed many of the relationships identified by SPM, more work is needed to evaluate the sensitivity and specificity of SPM relative to General Linear Model analyses.

CHAPTER VII

EXECUTIVE SUMMARY

Previous research examining the role of gluteal function in lower extremity control, particularly as it relates to functional valgus collapse, has yielded inconsistent results. These inconsistencies could in part be the result of methodological differences between studies (e.g. single-leg v. double-leg task evaluation, muscle strength v. muscle activation, lack of sex-stratified cohorts, and analyzing discrete time points v. the entire landing phase). Elucidating gluteal muscle influence on lower extremity biomechanics may be a critical step for the reduction of ACL injury rates, as neuromuscular dysfunction is likely more responsive to injury prevention efforts than are other risk factors such as bony anatomy, ligament quality, or hormonal influences, that are more difficult to modify. To that end, the overarching goal of this dissertation was to use more integrated analyses to comprehensively examine, within each sex, the role of gluteal strength and muscle activation *throughout* the load acceptance phase during both single and double leg landings, while also accounting for anatomical variations at the hip. We hypothesized that within each sex, functional valgus collapse, considered to be a movement pattern that increases one's risk for ACL injury, would be more pronounced during a single-leg landing task as opposed to a double-leg landing, and that once accounting for an individual's hip anatomy, the gluteal muscles would play an important role in controlling lower extremity motion, and potentially mediate any negative effects of anatomy on functional valgus collapse. If our hypotheses were correct, not only would it elucidate the importance of multifactorial and multivariate research designs moving forward, it would also identify specific areas in which interventions focused on neuromuscular training of the gluteal muscles might be most beneficial.

The results generally confirmed that the examination of lower extremity biomechanics during a single leg task was generally more telling than a double leg task in both sexes, though there were nuanced differences between sex which could have implications for injury risk screening and prevention. As expected, the single-leg landing task elicited more profound sex differences than did the double-leg landing task, particularly during the early stage of single-leg load acceptance when ACL injuries are thought to occur (30-40ms post initial ground contact). In addition to sex differences, the single-leg landing task resulted in greater functional valgus collapse, most notably knee abduction and hip adduction, which occurred consistent with the accepted timing of ACL injury. The gluteus medius also displayed greater activation during the single-leg task, suggesting that a single-leg task places greater demands on the hip musculature and may be more telling than a double leg task when screening individuals for risky hip and knee biomechanics.

However, even in the more demanding single leg landing, gluteal function did not mediate the relationship between femoral alignment and biomechanics. Rather, gluteal strength and activation explained a unique proportion of variance in lower extremity biomechanics beyond what was explained by femoral alignment. In females, weaker gluteal muscles predicted riskier frontal plane hip kinematics. In males, gluteal function was more associated with kinetics. This implies that our male cohort used their musculature to create torque about a joint, whereas our female cohort was unable to create torque, possibly resulting in compensatory movement. Overall, greater total hip ROM generally explained more variance in functional valgus collapse than did gluteal function, which was unexpected, as was the absence of any mediating effect by the gluteal muscles. Not only did total ROM explain considerably greater proportions of biomechanical variance, but observed associations between gluteal muscle function and biomechanics occurred 10-20ms after associations between femoral alignment and biomechanics.

Together, these suggest that the gluteal muscles are mechanically independent of femoral alignment. However, given the short time frame during which we observed these associations, gluteal muscle function could be temporally linked to one's femoral alignment. Even though these results demonstrated that femoral alignment may substantially impact lower extremity biomechanics, the varying mechanical and temporal relationships between function and anatomy could have significant implications for the screening and prevention of ACL injuries.

This study is expected to impact clinical practice as it pertains to screening and prevention of ACL injuries in athletic populations. Because total passive hip ROM was the strongest predictor of functional valgus collapse, obtaining both ROM_{IR} and ROM_{ER} may be useful for ACL injury screening. Secondly, gluteus medius and gluteus maximus strength testing could be a useful tactic to assess one's propensity for functional valgus collapse, as these were also predictors of poor biomechanics. Measuring both passive hip ROM and gluteal muscle strength would provide a more complete profile for ACL injury screening. When screening for high-risk biomechanics, sports medicine professionals should consider the biomechanical differences between single-leg and double-leg tasks, bearing in mind that the majority of ACL injuries occur during single-leg stance. Given the greater demands on the musculature to stabilize the knee upon weight acceptance, a single-leg task may be more adept than a double-leg task for identifying biomechanical patterns consistent with ACL injury. Further research could explore development of clinical criteria for standardization and quantification of a single-leg landing task, to include the role of trunk control, arm control, balance control, and functional valgus collapse; this would allow for easy identification of individuals at risk for displaying poor lower extremity biomechanics. Furthermore, the single-leg landing task elicited higher risk biomechanics during the time in the landing that is most consistent with what is reported for ACL injury. To that end, clinicians should consider examining high risk biomechanics throughout the landing phase, focusing on relevant time frames rather than peak values that are not time based. From an intervention standpoint, there is much work yet to be done. However, based on the current results, it may be useful to implement programs to strengthen the gluteal muscles and to encourage muscle pre-activation in individuals with excessive hip ROM to lessen their chances of sustaining ACL injuries as we continue to explore its functional role.

Although the current study suggested that the gluteal muscles might be a suitable target for screening and intervention, more research is needed to determine which aspects of gluteal function are modifiable and which intervention strategies are the most effective. While the current study used an untrained, randomly sampled cohort to demonstrate that gluteal function was mechanically independent yet temporally linked with femoral alignment, more research is needed to explore these relationships in cohorts of varying femoral alignment, as well as the possibility of altering gluteal function in these individuals. Specifically, if the observed temporal linkage between gluteal function and femoral alignment remains true in individuals with excessive ROM, this would suggest that greater muscle pre-activation in these individuals may not only be possible, but also beneficial. Further work should examine the degree to which alterations in gluteal muscle function can preclude the occurrence of functional valgus collapse. Due to the biomechanical differences between single-leg and double-leg landings, interventions addressing the gluteus medius may be may effective using a single-leg task. Though more research is needed in this area, the data in the current project suggests that individuals with weak gluteus medii also display excessive hip adduction, thus greater functional valgus collapse.

This study, though preliminary, provides a foundation from which future investigations can examine functional valgus collapse in a more integrated and holistic manner. Traditionally, the components of lower extremity biomechanical movement have been compartmentalized and analyzed independently of one another. While statistically convenient, this strategy fails to acknowledge inter-joint and inter-planar dependencies. Only recently are more advanced statistical tools being used to accommodate multidimensional data, and with the advent of analyses like SPM, multidimensional data can also be considered over time. Our studies revealed that while GLM was adequate for identifying and quantifying group differences and associative relationships, SPM was useful for identifying *when* in the landing phase significant group differences occurred. GLM and SPM were particularly complementary to one another in group analysis. The SPM CCA, which would be considered an associative, regression type analysis, was limited in its usefulness because of its inability to account for multiple predictors. This limitation resulted in a disconnect between SPM and GLM results and a cumbersome interpretation. Therefore, at the current time, researchers should limit SPM use to questions involving group comparisons, and exclude its use in multivariate correlative analysis. Taking advantage of the complementary nature of GLM and SPM in group analysis will allow for fuller and more complete interpretations. This study has also provided evidence that biomechanical effects consistent with high-risk functional valgus collapse occur early in the landing phase of a single-leg landing task. With this information, future work should at least choose pre-determined, discrete time points of interest for analysis, which would be an improvement over the accepted method of selecting peak position as the only time point of interest. Ideally, future work should seek out and employ analytical methods that allow for the inclusion of multiple independent variables, multiple dependent variables, and a time vector. Though it is not currently possible to account for all of these items in a single analysis, researchers should strive to conduct studies that mimic accepted injurious situations, and continually work toward the integration of time and multidimensional biomechanical movement.

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APPENDIX A

INTAKE QUESTIONNAIRES

PHYSICAL ACTIVITY AND HEALTH HISTORY

Do you have any General Health Problems or Illnesses? (e.g. diabetes, respiratory disease) Yes____ No____

Do you have any vestibular (inner ear) or balance disorders? Yes____ No____

Do you smoke? Yes____ No____

Do you drink alcohol? Yes____ No____ If yes, how often?__________________________

Do you have any history of connective tissue disease or disorders? (e.g. Ehlers-Danlos, Marfan's Syndrome, Rheumatoid Arthritis) Yes____ No____

Has a family member of yours ever been diagnosed with breast cancer? Yes____ No____ (if no, please skip next question.)

If yes, please put a check next to the types of relatives that have been diagnosed. You may check more than one box:

Mother________ Sister_______ Grandmother _______ Aunt ______.

Male relative (father, brother, grandfather, or uncle) _______.

Other type of relative (please write in) _________________________.

Please list any medications you take regularly: ______________________________

____

____

Please list any previous injuries to your lower extremities. Please include a description of the injury (e.g. ligament sprain, muscle strain), severity of the injury, date of the injury, and whether it was on the left or right side.

Body Part	Description	Severity	Date of Injury	L or R
Hip				
Thigh				
Knee				
Lower Leg				
Ankle				
Foot				

Please list any previous surgery to your lower extremities (Include a description of the surgery, the date of the surgery, and whether it was on the left or right side)

Body Part Description Date of Surgery L or R

____

____

____

____

Please list all physical activities that you are currently engaged in. For each activity, please indicate how much time you spend each week in this activity, the intensity of the activity (i.e. competitive or recreational) and for how long you have been regularly participating in the activity.

Activity #Days/week #Minutes/Day Intensity. Activity Began When?

____

____

____

____

____

____

____

What time of day do you generally engage in the above activities?_______________

____

____

Please list other conditions / concerns that you feel we should be aware of:_______

____

____

____

____

Please list any previous surgery to your lower extremities (Include a description of the surgery, the date of the surgery, and whether it was on the left or right side)

Body Part Description Date of Surgery L or R

Please list all physical activities that you are currently engaged in. For each activity, please indicate how much time you spend each week in this activity, the intensity of the activity (i.e. competitive or recreational) and for how long you have been regularly participating in the activity.

Activity #Days/week #Minutes/Day Intensity Experience in this Activity (# of years)

What time of day do you generally engage in the above activities?______________________

Please list other conditions / concerns that you feel we should be aware of:______________

KOOS KNEE SURVEY

Today's date: _____/______/______ Date of birth: _____/______/______

Name: ____

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities.
Answer every question by ticking the appropriate box, only <u>one</u> box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms
These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S3. Does your knee catch or hang up when moving?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S4. Can you straighten your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

S5. Can you bend your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

Stiffness
The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S7. How severe is your knee stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Pain

P1. How often do you experience knee pain?

Never	Monthly	Weekly	Daily	Always
☐	☐	☐	☐	☐

What amount of knee pain have you experienced the **last week** during the following activities?

P2. Twisting/pivoting on your knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P3. Straightening knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P4. Bending knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P5. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P6. Going up or down stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P7. At night while in bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P8. Sitting or lying

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P9. Standing upright

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A2. Ascending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A4. Standing

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A5. Bending to floor/pick up an object

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A14. Sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A17. Light domestic duties (cooking, dusting, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP2. Running

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP3. Jumping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP4. Twisting/pivoting on your injured knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP5. Kneeling

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never	Monthly	Weekly	Daily	Constantly
☐	☐	☐	☐	☐

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all	Mildly	Moderately	Severely	Totally
☐	☐	☐	☐	☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all	Mildly	Moderately	Severely	Extremely
☐	☐	☐	☐	☐

Q4. In general, how much difficulty do you have with your knee?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Thank you very much for completing all the questions in this questionnaire.

The Activity Rating Scale

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.					

Investigator Comments: