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Hip Mechanics of Unilateral Drop Landings

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HIP MECHANICS OF UNILATERAL DROP LANDING

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ABSTRACT

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Increased hip forces are a proposed factor for osteoarthritis and femoroacetabular impingement. These forces can be estimated through musculoskeletal modeling using measured kinematics and kinetics. An understanding of hip joint loading during landing in a asymptomatic population will begin to elucidate what, if any, sex differences exist and how changes in landing condition alter hip mechanics. The overall purpose of this dissertation was to explore how sex and landing condition effect landing mechanics. Landing mechanics were quantified using ground reaction forces (GRF), hip joint forces (HJF), and lower extremity kinematics during unilateral drop landings from 30-cm, 40-cm, and 50-cm, as well as, a 40-cm land-and-cut task. The relationships between sex and limb side, sex and landing task, and sex and landing height on landing mechanics were assessed using three sub-studies.

Eighty-three, recreationally active, adult volunteers completed landing tasks (40 participants completed the land-and-cut task). For sex-limb side, bilateral differences (right versus left) were examined at 40-cm. No bilateral differences were identified. For sex-landing task, 40-cm drop landings were compared to land-and-cuts. Higher peak GRF (pGRF) and pGRF loading rates were identified for landing-only. Landing-only tasks were performed with less ankle dorsiflexion range of motion for landing (ROM_L) and impact (ROM_I) phases. Landing-only tasks demonstrated more hip adduction ROM_L and more hip flexion ROM_I. For sex-landing height, landings were compared between 30-cm and 50-cm. Increasing landing height resulted in increased pGRF, pHJF, pGRF loading rate, and pHJF loading rate. With increased height, larger

3-D hip and knee flexion ROM_I and ROM_L were identified, as well as increased ankle dorsiflexion ROM_L.

There were no interaction effects between sex and landing condition. Sex differences across sub-studies demonstrated consistent trends. In all studies, females incurred larger pGRF compared to males, yet only the landing height analysis demonstrated increased pHJF. Females exhibited larger hip adduction and reduced hip rotation ROM_L. Females exhibited larger hip flexion, hip adduction, and knee flexion ROM_I. The landing task analysis identified increased female ankle dorsiflexion ROM_I.

Sex differences were identified between landing conditions, yet the lack of sex-landing condition interaction indicates both sexes may utilize similar modifications in response to changing landing conditions.

For my mom

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NOMENCLATURE

ACL	Anterior cruciate ligament
BW	Body weight
COM	Center of mass
dof	Degrees of freedom
F	Force (N)
FAI	Femoroacetabular impingement
GRF	Ground reaction force
HJF	Hip joint force
IC	Initial contact with the force plate
pGRF	Maximum resultant ground reaction force during landing
pHJF	Maximum resultant hip joint force during landing
m	Meter
MF	Maximum knee flexion during landing
MR	Magnetic resonance
OA	Osteoarthritis
PCSA	Physiological cross sectional area
ROM	Range of motion
ROM_I	Impact phase range of motion
ROM_L	Landing phase range of motion
s	Second
STD	Standard deviation

CHAPTER 1: INTRODUCTION

Femoroacetabular impingement (FAI) is the pathomechanical process created by anatomical abnormalities of the proximal femur and/or the acetabulum causing a mechanical abutment within the hip (Ganz et al., 2003; Ito et al., 2001; Leunig et al., 2009; Packer and Safran, 2015). Two classifications of bony impingement morphologies, cam and pincer, are associated with FAI. These subsets are classified by anatomical location (Ganz et al., 2003). Cam morphologies broadly describe a loss of sphericity of the femoral head, while pincer impingements are characterized by an over-coverage of the acetabulum (Bedi et al., 2008; Pfirrmann et al., 2006; Siebenrock et al., 2004). Cam and pincer bony morphologies may present in isolation; however, 50-70% of FAI diagnoses demonstrate some measure of both elements (Kapron et al., 2011).

Symptomatic FAI typically presents in young, active adults and adolescents as insidious groin pain worsened by long periods of sitting, walking, or general exercise (Bedi et al., 2008; Ganz et al., 2003; Leunig et al., 2006). The condition, when symptomatic, hampers hip range of motion (ROM) and imposes activity limitations, both in athletics and daily life (Diamond et al., 2014; Philippon et al., 2007b). Quality of life can be improved with arthroscopic correction (Malviya et al., 2012), however, the condition is difficult to diagnose. Early surgical intervention with symptomatic FAI increases the likelihood of long term success by reducing damage to the labrum and articular cartilage (Espinosa et al., 2006; Ito et al., 2001; Leunig et al., 2009). Unlike other arthritic conditions, FAI represents an “at risk” stage antecedent to developing end-stage hip arthritis.

Research on FAI has increased in recent years, yet the etiology is not well understood (Diamond et al., 2014; Packer and Safran, 2015; Reiman and Thorborg, 2015). Several etiological theories have been proposed including genetic factors, prior hip morphology (e.g. slipped capital femoral epiphyses), and athletic participation (de Silva et al., 2016; Packer and Safran, 2015). An association between high-level sports participation in adolescence and impingement morphology is rapidly gaining support (Agricola et al., 2012; Kapron et al., 2011; Siebenrock et al., 2011). Prevalence of FAI in adolescent and young adults varies to some degree by sport, with an increased incidence in cutting sports compared to flexibility, contact, impingement, or endurance based sports (Nawabi et al., 2014). In separate sport specific studies, athletes participating in high-levels of football, ice hockey, basketball, and soccer, beginning in childhood were at an increased risk of developing impingement morphology (Agricola et al., 2012; Gerhardt et al., 2012; Kapron et al., 2011; K. A. Siebenrock et al., 2011; Klaus A. Siebenrock, Kaschka, Frauchiger, Werlen, & Schwab, 2013). The combination of repetitive hip loading with flexion and internal rotation incurred while participating in cutting sports, may present physiologically excessive demands during sensitive adolescent skeletal development (Clohisy et al., 2013; de Silva et al., 2016; Siebenrock et al., 2011).

During this adolescent phase, skeletal development is more vulnerable to osseous adaptations in response to excessive and repetitive loading (Packer and Safran, 2015). Bone morphology remains highly susceptible to loading while the physis remains open and the individual is skeletally immature (Maffulli and Baxter-Jones, 1995; Nilsson et al., 2005). Longitudinal growth in the long bones occur at the physis, or growth plates. During adolescence, the physis environment is tremendously dynamic and lacks the resiliency of ligamentous and muscular structures (Mirtz et al., 2011; Nilsson et al., 2005). Perfusion within the femoral head can be impacted by repetitive loading, resulting in necrosis or stimulated bone growth (Maffulli and Baxter-Jones, 1995).

In addition to sport related risk factors, differences in the presentation and prevalence of symptomatic FAI have been linked with sex. Males are reportedly more likely to present with a cam or mixed morphologies (e.g. combination cam and pincer) and females more likely to present with pincer morphologies (Mascarenhas et al., 2016; Nepple et al., 2014b). Bilateral presentations are two to three time more prevalent in males compared to females (Klingenstein et al., 2013; Mascarenhas et al., 2016). Females self-reported reduced hip function and greater disability prior to hip arthroscopy (Nepple et al., 2014b). However, these sex differences were no longer identified after two years post-operation (Joseph et al., 2016). It is currently unclear how the proposed etiological factors contribute to sex-related differences in the development of symptomatic FAI.

Reductions in hip ROM are a key component in a symptomatic FAI diagnosis. The limited ROM is likely a combined result of anatomical incongruity, soft tissue edema, and pain avoidance strategies (Brisson et al., 2013; Diamond et al., 2014; Kennedy et al., 2009b). The aberrant morphology in FAI places physical constraints on joint function, wherein the joint simply cannot rotate through motion to the same extent as healthy hip (Ganz et al., 2003). In addition, soft tissue edema from recurrent impingement likely contributes additional movement constraints (Kennedy et al., 2009a). There are a limited number of three-dimensional (3-D) movement studies comparing symptomatic FAI with asymptomatic controls. Stair climbing was examined in one study using preoperative FAI candidates and controls (Rylander et al., 2013). As an ADL, stairs present a unique testing situation which requires a larger hip ROM and more muscular strength than walking (Bergmann et al., 2001). Results indicated a significant 6.2° reduction in hip sagittal ROM, due in large part to a significant 4.8° reduction in hip extension. Maximum internal rotation in the FAI group was significantly diminished by 5°, which is consistent with clinical testing FAI ROM data (Philippon et al., 2007b; Rylander et al., 2013). The combination of decreased internal rotation and extension may act as a pain avoidance strategy in attempting to avoid impingement with the femoral head and acetabulum (Kennedy et al., 2009a; Rylander et al., 2013). Drop landings were studied by Kumar et al. (2014) within the same study used to assess gait and deep squats. The 12-in drop landing height did not illuminate any kinematic or kinetic differences between FAI and healthy controls. Of note, FAI participants landed with their feet positioned more closely together. The smaller base of support may reduce the muscular strength needed to control the landing (Kumar et al., 2014). Movement studies have yet to illuminate clear kinetic trends in FAI mechanics potentially due to subject samples, small sample sizes, and task selection. Further, forces applied across the hip joint during dynamic tasks have not been studied. These forces can be estimated non-invasively through musculoskeletal modeling.

Estimation of hip joint forces began long before the age of computers. As early as the 1940s, researchers estimated hip forces during gait to fall somewhere in the vicinity of 4.4 to 4.6 times body weight (Rydell, 1966). Hip forces have traditionally been solved using an inverse dynamics approach with measured kinematics and ground reaction forces, then distributing the forces across load bearing structures surrounding the joint (Crowninshield et al., 1978). Optimization techniques have been extensively used to estimate joint forces. Static optimization has been criticized for oversimplifying system anatomy and failing to truly encompass muscle dynamics, yet it has been determined an effective method for force estimation in gait (Anderson and Pandy, 2001b; Wesseling et al., 2015).

Statement of the Problem

Diagnoses of symptomatic and non-symptomatic FAI are rapidly increasing (Colvin et al., 2012; Montgomery et al., 2013; Sampson, 2005). The incidence of impingement morphology is estimated as high as 10% to 15% of the general population and believed responsible for 22% to 55% of all hip pain (Groh and Herrera, 2009; Schilders et al., 2009). Enhanced radiographic detection presents a larger issue for those individuals with radiographic impingement evidence, yet no symptoms. In a recent review of literature analyzing over 2000 asymptomatic hips, cam morphology was observed in 37% and pincer morphology was observed in 67% (Frank et al., 2015). Currently, we have insufficient data to predict which hips are at an increased risk of developing hip OA (Kahlenberg et al., 2014). With current evidence, prophylactic surgical correction of impingement morphology is not indicated for non-symptomatic hips (Collins et al., 2014). Hip OA presents a major world risk, to which we do not have a cure nor a significant means of slowing disease progression (Felson, 2014; March et al., 2014).

It is possible to detect impingement morphology in athletes at a very young age, before significant damage has occurred. There is a strong likelihood many of these individuals will go on to develop symptomatic FAI, labral tears, and/or cartilage damage. Nonetheless, older adults have been reported to demonstrate similar morphologies that never progress into a symptomatic condition. Medical professionals need better explanations for the underlying etiological differences between symptomatic and non-symptomatic FAI in order to prescribe the most appropriate courses of treatment.

Increased forces between the femur and acetabulum are commonly accepted as a mechanism for damage in traditional hip OA, however it is unclear if FAI mechanics are related to increased hip force magnitudes, directions, or areas of force application (Ganz et al., 2003; Ito et al., 2001). Wolff's law has long been referenced to describe the ability of bone to undergo functional bone remodeling due to changes in its mechanical environment (Ruff et al., 2006). Considering the bony pathomorphology characteristic of FAI, it is reasonable to expect some degree of modified mechanical loading associated with the condition which elicits the altered bone structure. Unfortunately, hip contact forces cannot be directly measured via non-invasive measures. These forces can, however, be estimated through musculoskeletal modeling using measured kinematics and kinetics (Delp et al., 2007). These methods have been successfully used to estimate hip contact forces in normal and pathologic populations (Cleather et al., 2013; Modenese et al., 2011b). To date, hip contact forces in young, asymptomatic, active adults have not been examined during common sport actions such as, unilateral landing and cutting. These sport actions are prevalent amongst most sports and likely expose the hip to high forces. Developing an understanding of how asymptomatic hips respond to changes in landing limb side, landing height, and landing task will provide a baseline for examining potential changes associated with symptomatic hip pathologies.

Statement of Purpose

The overall purpose of this dissertation was to examine lower extremity mechanics, specifically at the hip, demonstrated by asymptomatic subjects performing unilateral drop landing and land-and-cut maneuvers. The landing mechanics were quantified using a combined approach of measured kinematics and kinetics, and musculoskeletal modeling.

Specific Aims and Hypotheses

<u>Specific Aim 1</u>: To identify sex differences in peak resultant hip joint force (pHJF), timing of pHJF, peak resultant ground reaction force (pGRF), timing of pGRF, average loading rates for pHJF and pGRF, as well as, lower extremity ROM during unilateral drop landings. When

significant differences were present in ROM, maximum and minimum joint angles were examined. Continuous HJF, GRF, as well as, 3-D hip, knee flexion, and ankle dorsiflexion measures were compared at each percentage of the landing.

Hypothesis: Males have been proposed to land with a more hip dominant landing strategy. As such, males were hypothesized to exhibit increased pHJF when compared to females. Females have been proposed to land with a stiffer landing configuration. As such, females were hypothesized to exhibit a diminished ROM in the hip, knee, and ankle and significantly larger normalized pGRF.

<u>Specific Aim 2</u>: To determine if sex and landing task influence pHJF, pGRF, timing of pHJF and pGRF, average loading rate for pHJF and pGRF, IC joint angles, and lower extremity ROM. Continuous HJF, GRF, as well as, 3-D hip, knee flexion, and ankle plantar/dorsi-flexion measures were compared at each percentage of landing.

Hypothesis: Land-and-cuts would incur larger peak forces and utilize less sagittal ROM compared to landing-only trials. It was also hypothesized females would incur larger peak force and utilize less sagittal ROM compared to males.

<u>Specific Aim 3</u>: To examine the relationships between sex and drop landing height with pHJF, pGRF, timing of pGRF and pHJF, average loading rate of pGRF and pHJF, lower extremity joint angles at IC, and lower extremity ROM. Continuous HJF, GRF, and 3-D hip, knee flexion, and ankle dorsi/plantar-flexion measures were compared at each percentage of the landing

Hypothesis: pHJF and pGRF would be significantly higher for 50-cm landings compared to 30-cm landings. Lower extremity ROM would increase for the hip, knee, and ankle measures when comparing the 30-cm landings to 50-cm landings. These differences would occur primarily in the sagittal plane. It was also hypothesized females would land with greater pHJF and pGRF compared to males with less sagittal ROM.

Limitations of the Study

1. The study sample represents a generalized asymptomatic population. The presence of abnormal hip anatomy is not known.
2. Retroreflective marker placement on anatomical landmarks is subject to human error.
3. Musculoskeletal models used are simple models and may not accurately depict individual joint movements. The hip is modeled with 3-dof which may provide more accurate information compared to the 1-dof models employed for the knee and ankle joints.
4. Several musculoskeletal model parameters, such as tendon slack length and physiological cross-sectional area, were not subject specific and may vary from participant to participant. The lack of subject-specificity of these model parameters may mask individual variances in estimating normal hip mechanics.
5. Muscle force in the musculoskeletal model was estimated using static optimization, which is not time dependent.

Delimitations of the Study

1. Participants were between the ages of 18 and 30.
2. Participants were in good general health and physically active. Physically active was defined as performing at least 30 minutes of exercise 3 times a week. At least one of these activity sessions was required to consist of repetitive jumping and landing, such as basketball or volleyball.
3. Participants had no history of lower extremity surgery or any health condition that affected their ability to jump.

4. Participants had not experienced an injury to the lower extremities in the past six months.
5. Participants wore lab standard shoes.

Assumptions of the Study

1. Participants provided honest information on all study documentation, including the health history questionnaire.
2. Joints were frictionless, segment masses were concentrated at the center of mass, and segments behaved as rigid bodies for inverse dynamics calculations.
3. The generic musculoskeletal model used in OpenSim fit the population utilized and adequately represented their true femoroacetabular kinematics.

Significance of the Study

FAI represents a significant and growing cause of disability in otherwise healthy active adolescents and adults. Advances in radiographic imaging and screening tools may allow for early detection of potentially detrimental morphology. Surgical intervention during this stage may slow or prevent degenerative hip arthritis from developing. With radiographic evidence observed in adolescents as young as twelve, there is a strong inclination to correct the “aberrant” anatomy in an effort to reduce joint damage risk. Hip arthroscopy was up 365% between 2004 and 2009, with the majority of patients between 20-yrs and 39-yrs of age (Montgomery et al., 2013). Colvin et al. (2012) reported an 18-fold increase in hip arthroscopy between 1999 and 2000. While short-term results and return to play statistics are optimistic, the long-term impact on development of hip OA remains to be seen.

The current available information pertaining to FAI leaves us in a conundrum. Should healthcare providers act quickly to correct aberrant morphology upon detection, before permanent joint damage has occurred? At what stage of symptomatic progression is it appropriate to consider surgical treatment? Young adult and adolescent hip pain present complicated issues in treatment and symptom management with the potential for significant financial costs and lifestyle limitations. With a strong link emerging between athletic participation and impingement morphology, we need a comprehensive understanding of asymptomatic hip mechanics during dynamic, sport related tasks. Data obtained through this research may improve understanding of hip biomechanics during landing, which can be used as a baseline for future studies with a symptomatic cohort. The multifaceted research approach seeks to identify changes during landing with respect to kinematics, kinetics, and hip joint contact force. The ramifications of these findings are pertinent within the context of pathological hip conditions.

Operational Definition of Terms

<u>Femoroacetabular impingement</u>: Pathomechanical condition of the femoroacetabular joint resulting in abnormal joint function.

<u>Femoroacetabular joint</u>: Ball and socket joint formed by the acetabulum and proximal femur.

<u>Hip Joint Force</u>: The force transferred from the femur to the pelvis.

<u>Kinematics</u>: Mechanical analysis of the motion of an object.

<u>Kinetics</u>: Mechanical analysis of the forces acting on an object.

<u>Musculoskeletal modeling</u>: The use of mechanics to model human motion with respect the muscles and bones.

CHAPTER 2: REVIEW OF LITERATURE

Introduction

The review of literature will have three primary areas of focus: femoroacetabular impingement, mechanics of landings, and musculoskeletal modeling. First, a discussion on the epidemiology and mechanics associated with FAI will lay a foundation with which to examine the need for understanding hip mechanics in sporting activity. Then, a review of our current understanding on lower extremity mechanics during landing will lend support and offer a foundation for a sound methodological design for studying landing hip joint mechanics. Finally, a summary of musculoskeletal modeling will describe the basis for key assumptions and mathematical relationships necessary for analysis of hip specific kinetic variables during landing. The integration of these three components will fully support the need for hip landing mechanics research, the basis for proposed methodology, and justify the use of musculoskeletal modeling.

Femoroacetabular Impingement

One of the earliest reports of hip impingement emerged as a secondary outcome in patients undergoing periacetabular osteotomy (Myers et al., 1999). The goal of periacetabular osteotomy is to ameliorate congenital hip dysplasia and reorient the acetabulum to correct insufficient femoral head coverage. However, post-surgical complaints of groin pain and limited internal rotation persisted despite the corrected acetabular coverage. Upon further examination, Myers et al. (1999) identified a secondary impingement caused by a geometric incongruity of the femoral head-neck junction and the newly oriented acetabulum. An additional surgery to reshape the femoral head-neck junction, deepening the offset, was successful in restoring flexion and internal rotation without boney impingement. Hence, surgically correcting the impingement between the femur and acetabulum.

Within two years of the Myers et al. (1999) study, reports describing the presence of similar femoroacetabular impingements in non-dysplastic hips appeared (Eijer et al., 2001b; Ito et al., 2001). Today a Web of Science search with the key words, “femoroacetabular impingement” returns over 2,000 entries. FAI is now recognized a significant source of hip pain, resulting in limitations during activities of daily life. Surgical correction for FAI is deemed “medically necessary” and a covered benefit of most major health insurance companies. Understanding of the condition has grown volumes, yet there is a great deal we do not understand with respect to etiology and predictors of deterioration to hip OA.

Epidemiology

Put simply, FAI is a condition brought about by a lack of fit between the head of the femur and the acetabulum. Anatomically, there is insufficient clearance between the bony structures to allow normal movement without creating bone-on-bone contact. The aberrant anatomical morphologies responsible for impingement are specific to the femur (cam) and acetabulum (pincer) (Appendix A).

Cam morphologies are characterized by a loss of sphericity of the femoral head and decreased femoral neck offset (Ito et al., 2001). Loss of sphericity may be brought about by an osseous deformity, known a pistol grip, which increases femoral head radius where it should begin to taper into the femoral neck (Stulberg et al., 1975). As the malformed femoral head rotates within the acetabulum, a cam-effect occurs at hip internal and abductionROM boundaries. At this point, the femoral head’s rotational motion is translated to abnormal linear motion resulting in shear and compressive forces being transmitted to the acetabulum, articular cartilage, and labrum. Shear forces incurred as a result of this motion are detrimental to the labrum and acetabular cartilage. With cam morphology, acetabular cartilage may undergo an outside-in wear pattern and potentially lead to avulsion from the labrum (Ganz et al., 2003).

Pincer morphologies are characterized by an over coverage of the acetabulum which creates an excessively deep socket (Ganz et al., 2003). Both, coxa profunda (i.e. general over coverage) and acetabular retroversion (i.e. local over coverage) limit ROM as the femoral-neck junction contacts the acetabular rim. Under these conditions, a linear force is applied to the acetabular rim with each contact. As a result, excessive forces may lead to labral damage and potentially degenerative changes. Acetabular rim ossification can occur over time, compounding the issue due the additional increase in acetabular depth (Ganz et al., 2003).

Cam and pincer morphologies often present in conjunction, to some degree. Radiographic evidence of both morphologies can be identified in 50 to 80% of impingement diagnoses (Ganz et al., 2008; Ganz et al., 2003; Kapron et al., 2011). In mixed cases, characteristics of both morphologies' aberrant loading characteristics are hypothesized to occur. To this end, mixed morphology FAI may present with intra-articular damage to the labrum, articular cartilage, and acetabulum (Ganz et al., 2008). Conversely, some researchers have proposed the two morphologies result in entirely different conditions (Cobb et al., 2010; Laborie et al., 2011).

Prevalence. The number of patients diagnosed with FAI has risen significantly in the past decade (Griffin et al., 2016), paralleling an increased diagnostic rate of hip injuries over the same time period (Orchard, 2015). It is estimated 20 million Americans have symptomatic FAI (Collins et al., 2014; Ganz et al., 2003). A clear distinction between symptomatic and asymptomatic FAI must be established before discussion of FAI prevalence. In 2016, the Warwick Agreement on FAI released a statement proposing FAI syndrome be used when referring to the combination of impingement pathomorphology, clinical indications and symptoms (Griffin et al., 2016). To date, widespread acceptance of the terminology FAI syndrome has not occurred. Asymptomatic FAI refers to individuals with radiographic impingement morphology, but do not have reported symptoms and/or clinical indications. Several studies assessing prevalence of impingement morphology in symptomatic and asymptomatic individuals have been performed (Gosvig et al., 2008; Hack et al., 2010; Heijboer et al., 2014; Kapron et al., 2011; Laborie et al., 2011; Philippon et al., 2013; Pollard et al., 2010c; Siebenrock et al., 2011; Siebenrock et al., 2013). Current studies demonstrate several reoccurring descriptors associated with prevalence of impingement morphology: age, sex, adolescent sport participation, and specific sport participation. Due to study differences in definition of impingement morphology (i.e. alpha angle threshold), clinical examination, or presence of symptoms direct comparisons between studies are challenging. The discussion will address the prevalence of symptomatic and asymptomatic impingement morphology.

The prevalence of impingement morphology, particularly cam, is assessed most often in terms of alpha angle (Nötzli et al., 2002). Alpha angles are traditionally measured using axial imaging to quantify the geometry of the femoral head and neck using the two lines. The first line is placed through the geometric center of the femoral head and through the midline of the femoral neck. The second line is placed where the femoral head neck junction contour goes beyond the femoral head radius (Nötzli et al., 2002). Alpha angles associated with cam impingement vary by study and range from 50° to 83°, with a majority of researchers using values between 50° and 55° (Nötzli et al., 2002). Going forward, variations in study reports of cam prevalence will have inherent disjunction based solely on cam diagnostic criteria (Griffin et al., 2016) and populations examined.

In the general population, the prevalence of impingement morphology was originally estimated between 10% and 15%; however, in more recent studies a wider range of values are believed to exist (Gosvig et al., 2008; Laborie et al., 2011). A prospective study examining the hips of young adults (mean age = 18.6-yrs) presenting for conscription into the Swiss military demonstrated asymptomatic radiographic morphologies in 6.1 to 21% of participants (Laborie et al., 2011). This study benefitted from a large sample size (n=2081); however, researchers did not include alpha angle. While alpha angle has been argued as a flawed method for diagnosis, it is readily used and allows for comparison between studies. Another large population study (n=4151) was conducted in conjunction with the Copenhagen Osteoarthritis Sub study that detected cam impingements in 17% of males and 4% of females (Gosvig et al., 2008). The study sample was significantly older with a mean age of 59-yrs and 60-yrs for women and men respectively. Further, the prevalence may be under representative due to extremely high alpha angle thresholds of 83° for males and 57° for females.

The prevalence of impingement morphology with respect to sex is currently undetermined. The preponderance of research has proposed young adult males are more prone to cam morphology, while athletic middle aged women are more likely to incur pincer morphology (Beck et al., 2005; Clohisy et al., 2010; Ganz et al., 2008; Ganz et al., 2003; Hack et al., 2010; Ito et al., 2001). Conflicting findings have indicated similar numbers of pincer morphology between sexes (Laborie et al., 2011).

When considering all impingement morphologies, between 20% and 75% present with radiographic evidence of impingement morphology bilaterally (Allen et al., 2009; Ganz et al., 2008; Klingenstein et al., 2013). Bilateral presentations may present as asymptomatic, have only one symptomatic hip, or with bilateral symptoms. When presenting with one symptomatic side, researchers were unable to identify the symptomatic side based off bilateral radiographs (Klingenstein et al., 2013). However, when bilateral morphology is present, higher alpha angles were observed than in unilateral morphological presentations (Allen et al., 2009). Possible underlying genetic factors (e.g. histobiological or joint structure) may be linked with the high prevalence of bilateral presentation and the associated increased alpha angle (Pollard et al., 2010c).

Level of sport participation is associated with the prevalence of impingement morphology (Agricola et al., 2012; de Silva et al., 2016; Heijboer et al., 2014; Kapron et al., 2011; Kapron et al., 2012; Kapron et al., 2015; Larson et al., 2013; Philippon et al., 2013; Siebenrock et al., 2011; Siebenrock et al., 2013). When considering elite level athletes, the prevalence of one radiographic measure of FAI has been estimated as high as 95% (Kapron et al., 2011; Weir, 2013). This increased risk of impingement morphology may be due to abnormally high training loads and volumes (Agricola et al., 2012; Keogh and Batt, 2008). Athletes with participation in practice at least three days a week were more likely than non-athletes to have one indicator of impingement radiography (de Silva et al., 2016). The preponderance of surveillance data is available only for elite and near elite level athletes. Unfortunately, this leaves a void when recreational athletes are considered (Orchard, 2015). Daily activity level and impingement morphology was tracked in a cohort study of adolescents (Carsen et al., 2014). Findings demonstrated an increased activity level was significantly related with the presence of impingement morphology.

The incidence of impingement morphology varies by sport. According to a meta-analysis, male adolescent athletes participating in ice hockey, basketball, and jumping sports in general were at an increased relative risk of developing abnormal morphology compared with controls (de Silva et al., 2016; Nepple et al., 2015). In collegiate football, upwards of 70% of players have been identified with alpha angles greater than 55° and 95% have at least one sign of impingement (Kapron et al., 2011; Larson et al., 2013). Elite adolescent basketball players demonstrated an 89% occurrence of alpha angle greater than 55° compared to 9% of aged matched controls (Siebenrock et al., 2011). Adolescent hockey players' alpha angles were significantly higher when compared to skiers. With an alpha angle of 55° as the threshold, 75% of hockey players had evidence of impingement compared to only 42% of skiers (Philippon et al., 2013). Research trends are beginning to show an increased risk related with specific sports and types of sports. However, these associations may be confounded in relation to body types, training habits, and lifestyle choices common to the sport.

The prevalence of impingement morphology, specifically alpha angle, has been tracked with respect to physeal closure (Heijboer et al., 2014; Philippon et al., 2013; Siebenrock et al., 2011; Siebenrock et al., 2013). Philippon et al. (2013) found an 18% increase in impingement morphology for adolescent hockey players when comparing open physis to closed physis, with an increased alpha angle after growth was complete. Siebenrock et al. (2013) also examined adolescent hockey players and reported an increase in alpha angle from the normal physiological range with an open physis to an alpha angle greater than 55° after physis closure. Similarly, impingement morphologies increased from 7% to 22% when adolescent soccer players were tracked for two years (Heijboer et al., 2014). Increases in age and level of play likely parallel increasing alpha angles over time (Philippon et al., 2013).

Demographic and sport data of surgical FAI arthroscopy candidates indicated high-level athletes preparing to undergo arthroscopic FAI correction were significantly younger and more often male than recreationally active candidates (Nawabi, Bedi, Tibor, Magennis, & Kelly, 2014). In addition, these high-level athletes were more likely to have participated in cutting sports than flexibility, contact, impingement, or endurance based sports. Most alarming, the mean patient age at time of surgery was 27-yrs, with a majority falling between 15-yrs and 22-yrs (Nawabi et al., 2014).

Presentation and diagnosis. Presentation of symptomatic FAI initiates as a groin pain aggravated with prolonged sitting or standing and exercise (Ganz et al., 2003; Philippon et al., 2007a). The onset of symptoms is typically insidious, yet patients may recall a minor trauma as the precursor to symptoms. Patients often demonstrate the “C” sign, cupping the hand around the hip just above the greater trochanter when asked to indicate where the pain is felt (Dooley, 2008). Presentation is generally described as a gradual onset of hip pain, which occasionally may radiate laterally toward the trochanteric region, medially into the adductor region, and rarely into the buttocks or even down to the knee (Griffin et al., 2016; Philippon et al., 2007a; Sankar et al., 2013).

Common findings upon clinical examination are a positive impingement test and decreased hip internal rotation (Ganz et al., 2003; Nötzli et al., 2002). An impingement test is performed with the patient supine on a firm surface then flexing, adducting, and internally rotating the hip. In normal hips, the combined motion places the femoral head neck junction and acetabular rim in close proximity, while physical contact between structures occurs with symptomatic FAI (Ganz et al., 2003). In symptomatic FAI, pain is elicited when an additional force is applied towards internal rotation due to shearing forces within the potentially inflamed and/or damaged intra-capsular structures (Ganz et al., 2003; Leunig et al., 2000). Hip internal ROM is evaluated with the patient supine and hip and knee flexed to 90° (Nötzli et al., 2002). Manual or electric goniometers are used to gauge the ROM from neutral to terminal motion

(Kapron et al., 2012). Healthy hips fall within 30-45° ROM, and values below this threshold could considered limited (Roach and Miles, 1991). Pre-surgical FAI patients have been documented with internal rotation ROMs from 7-15°, which is a large deviation from the normal range (Eijer et al., 2001a; Kubiak-Langer et al., 2007; Lavigne et al., 2004; Siebenrock et al., 2003). Clinical evaluations cannot positively identify hip joint pathologies; however, they can rule out those individuals whose symptoms are not due to a hip pathology (Reiman and Thorborg, 2015).

Subjective measures of hip health include standardized and generalized self-reported assessments. Generalized assessments are unique to institutions (e.g. clinics or universities) and are used to gather health history information and injury specific details. Such details include, demographics, descriptors of symptom onset, duration of symptoms, aggravating and alleviating factors, and lifestyle and activity characteristics. This data is used to paint a general description of the patient’s history and lifestyle. Standardized assessments include the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) (Bellamy et al., 1988), hip outcome score (Martin et al., 2006), and Harris hip score (Harris, 1969).

When physical and subjective examinations warrant further investigation, radiographic and pharmacological diagnostic tools are available. Intra-articular hip injections, may be given to further support intra-articular morbidities (Ayeni et al., 2014). Injections containing an anesthetic and a corticosteroid serve a dual role as a therapeutic modality and diagnostic tool. The combination of medications reduce inflammation while relieving pain originating within the hip capsule (Ayeni et al., 2014). Significant pain relief after injection lends support to an intra-articular mechanism as the underlying cause of pain as opposed to referred pain from a secondary condition (Ayeni et al., 2014; Klingenstein et al., 2013). Pain relief is not a strong predictor of post-surgical outcome; however, lack of pain relief is a good indicator surgical correction will not improve symptoms (Ayeni et al., 2014).

Radiographic diagnosis is an essential component for arriving at an FAI diagnosis. Radiographs, magnetic resonance imaging (MRI), magnetic resonance arthrography (MRA), and commuted tomography (CT) may be used to identify soft tissue and bony comorbidities (Clohisy et al., 2007; Tannast et al., 2007a; Tannast et al., 2007b). Physical findings suggestive of cam morphology include an osseous bump on the femoral neck (e.g. pistol grip deformity), fibrocystic changes at the femoral head neck junction, and acetabular rim edema (Register et al., 2012). In addition, radiographic examination includes visualization of potential soft tissue injuries including labral tears, loose bodies, and OA.

The relationship between labral tears and FAI is complex. When examining patients with labral tears, 49-87% were found to demonstrate at least one bony morphology consistent with FAI (Guevara et al., 2006; Peelle et al., 2005; Wenger et al., 2004). Like symptomatic FAI, labral tears are associated with an increased risk of osteoarthritis (Peelle et al., 2005). In a blind review of asymptomatic patients, 73% demonstrated some abnormal morphology and 69% had evidence of a labral tear (Register et al., 2012). Presence of a labral tear does not directly indicate underlying impingement morphology; however, when considered with additional radiographic evidence it increases the likelihood of a FAI diagnosis.

To date, a consensus for the definitive radiographic evidence needed to diagnosis FAI has not been reached (Chaudhry and Ayeni, 2014; de Silva et al., 2016). Quantitative measures from radiographic images provide a more objective means for diagnosis compared to using clinical and self-reported tools alone. Generally accepted ranges for alpha angle (Nötzli et al., 2002), internal rotation ROM, lateral center edge angle (Wilberg, 1939), head neck offset (Eijer et al., 2001a), acetabular index, and femoral anteversion are available. Specific techniques vary with imaging parameters and patient position; however, the measurement concepts remain similar.

Femoral morphologies are most commonly measured radiographically by alpha angle and head neck offset or ratio (Eijer et al., 2001a; Myers et al., 1999; Nepple et al., 2014a). Alpha angle is a quantification of femoral neck concavity, with larger alpha angles indicative of diminished head neck concavity (Nötzli et al., 2002). It is a quantitative representation of the osseous bump associated with cam impingement. Alpha angle measurements demonstrate excellent intra-observer reliability, but only fair to moderate inter-observer reliability (Carlisle et al., 2011; Nepple et al., 2014a). An incongruity exists for the diagnostic threshold as well, with pathological values ranging from 50° to 83° across physicians and researchers (Nepple et al., 2014a; Nötzli et al., 2002; Reiman and Thorborg, 2015; Tannast et al., 2007b). To further complicate matters, individual variation in the general population has reported alpha angles ranging from 32° to 62°, indicating a wide range of “normal” (Pollard et al., 2010c). On the positive side, alpha angles can be measured relatively quickly using standard x-ray imaging techniques which allows use across heath care and research institutions. Alpha angles can easily be compared, yet they must be interpreted with caution.

Pincer morphology can be measured radiographically by the cross over sign (COS) and lateral center edge angle (LCEA) (Tannast et al., 2007b). Measurements of acetabular depth have been utilized in the investigation of hip dysplasia and demonstrate good reliability (Carlisle et al., 2011; Mast et al., 2004; Tönnis and Heinecke, 1999). The LCEA is a measure of lateral acetabular coverage based on measurements (Wilberg, 1939). Values less than 20 are indicative of an over-coverage, characteristic of a pincer impingement (Tönnis and Heinecke, 1999). COS is a descriptor of acetabular retroversion in FAI, in which the anterior acetabular rim extends too far laterally to the medial border (Tönnis and Heinecke, 1999). The test is considered symptomatic of abnormal morphology when positive.

Variations in imaging tools, patient positioning, and radiographic measurement techniques create challenges in comparing findings between physicians, clinics, and studies (Carlisle et al., 2011). The complexities surrounding symptomatic versus non-symptomatic FAI illustrate the importance of a clearly defined, widely accepted criteria for diagnosis. According to Sankar et al. (2013), the Medical Subject Headings Thesaurus of the US National Library of Medicine provides five guidelines for the diagnosis of femoroacetabular impingement, "(1) abnormal morphology of the femur and/or acetabulum, (2) abnormal contact between these two structures, (3) especially vigorous supraphysiologic motion that results in such abnormal contact and collision, (4) repetitive motion resulting in the continuous insult, and (5) the presence of soft-tissue damage." It is important to note no mention of symptoms are required for diagnosis (Reiman and Thorborg, 2015; Sankar et al., 2013). In 2016, a symposium was convened to standardized FAI language, diagnostic measures, and criteria (Griffin et al., 2016). Going forward, the symposium suggested the addition of patient reported symptoms and clinical assessment be co-requisites for FAI syndrome diagnosis.

Treatment. Conservative and surgical treatment options for symptomatic FAI exist (Lavigne et al., 2004). Following a FAI diagnosis, orthopedic surgeons may require a period of conservative treatment before considering surgical courses of action (Espinosa et al., 2006; Philippon et al., 2007c). Opinions vary on the appropriateness of conservative treatment in fully developed FAI, as prolonging the time before surgical correction may increase the potential for irreversible intraarticular damage.

Conservative treatments include the use of drugs for pain and inflammation management, intraarticular injections, physical therapy, and activity modification (Feeley et al., 2008; Klingenstein et al., 2013; Lavigne et al., 2004). Oral medications may be prescribed, including non-steroidal anti-inflammatory drugs (NSAIDs) and narcotics to ameliorate pain (Emara et al., 2011). Physical therapy protocols are targeted on relieving inflammation, maintaining proper muscular control, and adapting activity modifications to prevent additional damage (Emara et al., 2011). It is important to note, conservative treatment cannot correct the underlying abnormal morphology and thus cannot eliminate FAI (Ganz et al., 2003). Due to the mechanical nature of the condition, the only way to prevent further joint damage using conservative treatment is through life long activity modification (Emara et al., 2011; Lamontagne et al., 2009b).

The goals of surgical treatments are to restore normal bony morphology and ROM, decrease pain, and improve long term joint heath (Kelly et al., 2012). Open, arthroscopic, and combined surgical approaches have been utilized to successfully restore natural joint motion (Clohisy et al., 2010; Ganz et al., 2003; Laude et al., 2009; Philippon et al., 2007d). Contraindications to surgery include hip dysplasia, posterior extension of cam morphology, and osteoarthritis (i.e. greater than Tonnis grade 1) (Beck et al., 2005; Klingenstein et al., 2013). Murphy et al. (2004) found a negative relationship between level of osteoarthritic change and surgical outcomes.

Specific surgical treatment procedures include acetabular rim trimming, labral debridement, labral refixation, and femoral osteochondroplasty (Byrd and Jones, 2011). When labral tears are present, debridement and refixation are the preferred course of treatment (Espinosa et al., 2007). However, with extensive damage labral resection may be required.

Surgical treatment carries an inherent risk. The risk is considered justifiable when managing pain and preventing additional joint damage associated with symptomatic FAI. Many cases of impingement morphology deteriorate into early OA (Audenaert et al., 2011; Bardakos and Villar, 2009). Asymptomatic FAI presents a different perspective for assessing surgical risks. Currently, prophylactic surgical treatment for asymptomatic FAI is not indicated based on a lack of long-term prospective studies providing evidence for progression to joint degeneration (Collins et al., 2014).

Etiology. The underlying factors leading to impingement morphology and FAI syndrome remain ambiguous. Proposed factors include pre-existing morphologies, genetic factors, and early, high-level athletic exposure (Chaudhry and Ayeni, 2014; Leunig et al., 2009). A true causal relationship between these factors and FAI syndrome has not been determined due to a lack of long-term prospective studies. However, early evidence indicates the causes are likely multifactorial and due to an interaction between lifestyle, biology, and potentially trauma (Klingenstein et al., 2013).

Pre-existing hip morphologies are an accepted casual factor in impingement morphology development (Fraitzl et al., 2007; Ganz et al., 2008). A variety of hip morphologies, such as hip dysplasia, SCFE, and Legg-Calve-Perthes, have been extensively examined and directly associated with OA development (Hoaglund and Steinbach, 2001; Jacobsen and Sonne-Holm, 2005; Leunig et al., 2000; Murphy et al., 2004). In these conditions, a gross deformity results in an improper fit between the femoral head and acetabulum which is attributed to joint degeneration (Ganz et al., 2008; Leunig et al., 2000). In these cases, a subtle impingement morphology may present due to the preexisting mechanical environment, further inhibiting natural joint motion (Leunig et al., 2000; Rab, 1999). The presentation and progression of impingement morphologies in conjunction with concomitant hip disease is difficult to generalize based on the large number of confounding variables. With current research, we can only definitively report that they co-exist and are interrelated.

The role of genetics as a causal factor in impingement morphology has been proposed (Pollard et al., 2010b). In the Pollard et al. (2010b) study, FAI patients were compared with their siblings while using their spouses as controls. They observed the siblings were at a 2.0 relative risk of having a cam morphology and a 2.8 risk of having a pincer morphology when the morphology was present in the patient. In addition, the prevalence of impingement pathomorphology was significantly higher in the siblings compared to spouse controls. According to their findings, modified hip anatomy may be due to a genetic disposition or determined at conception. Genetic factors have been associated with comparable acetabular pathomorphologies (e.g. hip dysplasia), while femoral pathomorphologies (e.g. SCFE) have not demonstrated a genetic link (Pollard et al., 2010b). Hence, it is possible genetic factors and/or histology play a role in the development of symptomatic FAI. To date, only one study has examined this relationship. Limitations in study design do not account for compounding factors such as similar lifestyles in siblings during development compared to potentially different developmental lifestyles of spouses (Chaudhry and Ayeni, 2014).

The role of adolescent lifestyle as an etiological factor in the development of impingement morphology is rapidly gaining support. In the early 1970's, Murray and Duncan (1971) proposed a relationship between school sports participation and a "femoral head tilt deformity." Enhancements in radiography and clinical awareness of hip conditions have promoted early detection of hip pathologies (de Silva et al., 2016). Recent findings indicate impingement morphology begins to form as early as 10-yrs to 12-yrs (Monazzam et al., 2013).

During this adolescent phase, skeletal development is more vulnerable to osseous adaptations in response to excessive and repetitive loading (Packer and Safran, 2015). Bone morphology remains highly susceptible to loading while the physis remains open and the individual is skeletally immature (Maffulli and Baxter-Jones, 1995; Nilsson et al., 2005). Longitudinal growth in the long bones occur at the physis, or growth plates. During adolescence, the physis environment is tremendously dynamic and lacks the resiliency of ligamentous and muscular structures (Mirtz et al., 2011; Nilsson et al., 2005). Perfusion within the femoral head can be impacted by repetitive loading, resulting in necrosis or stimulated bone growth (Maffulli and Baxter-Jones, 1995).

The proximal femoral physis is located at the proximal end of the femoral head neck junction, adjacent to where the osseous bump occurs in impingement morphologies. It has been proposed that increased exercise volumes and intensities at a skeletally immature age are supraphysiologic, resulting in abnormal bone formation at the head neck junction. On average, physis fusion occurs between the ages of 15-yrs and 22-yrs (Buikstra and Ubelaker, 1994). After adolescence, once physis closure is complete, the same volume and load may become more manageable. After the proximal femoral physis has closed, the femoral osseous bump does not appear to arise. However, once morphology is present, alpha angle demonstrates a tendency to increase with age. It is possible that abnormal, reoccurring loads are incurred as a result of the initial impingement morphology. Thus, creating a perpetual cycle of degenerative mechanics. Additional bone growth can occur well past the adolescent years, further increasing the size of the initial morphology with repeated loading.

Symptomatic FAI Mechanics

Reductions in hip ROM are a key component in a symptomatic FAI diagnosis. The limited ROM is likely a combined result of anatomical incongruity, soft tissue edema, and pain avoidance strategies (Brisson et al., 2013; Diamond et al., 2014; Kennedy et al., 2009b). The aberrant morphology in FAI places physical constraints on joint function, wherein the joint simply cannot rotate through motion to the same extent as healthy hip (Ganz et al., 2003). In addition, soft tissue edema from recurrent impingement likely contributes additional movement constraints (Kennedy et al., 2009a). Outside of the hip, we must consider the impact of pain and pain avoidance on neuromuscular movement strategies (Beaulé et al., 2007; Kennedy et al., 2009b; Rylander et al., 2011). All examinations of symptomatic FAI will be limited in their ability to quantify mechanics modifications specific to the underlying pathomorphology due to the individual compensatory strategies utilized to protect the hip. Conversely, we do not know which asymptomatic hips will progress into symptomatic FAI. Thus, without prospective studies we must proceed with caution in interpreting results. The mechanics associated with gait, squatting, and stair climbing have been examined and illustrate modifications associated with symptomatic FAI.

Prolonged walking is frequently reported as a symptom inducing activity for individuals with symptomatic FAI. Several studies have used three-dimensional movement analysis to examine hip kinematics and kinetics associated with level walking in pre-surgical FAI patients (Brisson et al., 2013; Diamond et al., 2014; Hunt et al., 2013; Kennedy et al., 2009b; Rylander et al., 2011; Rylander et al., 2013). There were no spatiotemporal differences observed in all studies when compared to either healthy controls or post-surgical results. Symptomatic FAI consistently demonstrated a significant reduced sagittal hip ROM of approximately 4° in several studies (Brisson et al., 2013; Diamond et al., 2014; Hunt et al., 2013; Rylander et al., 2011). There was a mixed consensus as to the underlying movement direction, with two studies reporting limited hip flexion and one finding limited extension. Movement in the transverse and frontal planes did not yield as homogenous results. In the transverse plane, Hunt et al. (2013) reported a reduction peak internal rotation while Rylander et al. (2013) reported reduced internal rotation ROM. It should be noted, transverse hip motion has characteristically been the most difficult to capture accurately and with good repeatability (Besier et al., 2003). The remaining gait studies mentioned did not identify differences associated with the transverse plane. The frontal plane analyses yielded similarly confounding results. Only Brisson et al. (2013) and Kennedy et al. (2009a) found significant reductions in frontal ROM, including a reduced peak hip abduction angle after toe-off.

Kinetic differences in hip joint moments were examined in four of the studies using pre-surgical patients and controls (Brisson et al., 2013; Diamond et al., 2014; Hunt et al., 2013; Kennedy et al., 2009a). In all but one study, no significant differences in any of the joint moments were reported. Hunt et al. (2013) reported reductions in both the hip flexor and internal rotation moments. These deficits were suspected to be related to reduced muscular activations in early and late stance, but could not be confirmed without EMG data.

The most consistent finding in gait analysis of symptomatic FAI is a reduction in sagittal plane motion. Hip flexion in symptomatic FAI when compared to healthy controls is associated with a larger angular deficit than abduction and internal rotation (Nussbaumer et al., 2010). The 3-D hip ROM utilized during gait should fall within the impingement free ROM available to individuals with symptomatic FAI. However, when considering additional constraints from soft tissue, perhaps the limited sagittal ROM available approaches those ranges experienced in gait.

As previously discussed, modified movement strategies may be used as a protective mechanism. It also important to note that reported differences in hip kinematics for all planes were relatively small, between 3° and 5°(Brisson et al., 2013; Hunt et al., 2013; Rylander et al., 2013). Although these small changes were associated with statistically significant findings, the clinical impact may be negligible (Diamond et al., 2014).

Additional motional analysis studies have been performed comparing symptomatic FAI and healthy controls for squatting (Bagwell et al., 2016; Kumar et al., 2014; Lamontagne et al., 2009b), stair climbing (Kumar et al., 2014; Rylander et al., 2013), and drop landings (Kumar et al., 2014). The large ROM used during these movements enables a better assessment of lower limb mobility associated with symptomatic FAI (Konishi and Mieno, 1993). Additionally, squatting and stair climbing represent activities of daily life (Lamontagne et al., 2009b). Pre-surgical FAI groups demonstrated reduced squat depth when compared to healthy controls (Bagwell et al., 2016; Lamontagne et al., 2009b). Modified three-dimensional kinetics and kinematics were identified between groups as well.

In the sagittal plane, squat kinematics consistently demonstrated no significant differences in hip ROM (Bagwell et al., 2016; Kumar et al., 2014; Lamontagne et al., 2009b). However, differences in pelvic motion were identified in the FAI groups. With FAI, Bagwell et al. (2016) identified a reduction in pelvic tilt when lowering into the squat, which resulted in a more anteriorly positioned pelvis at maximum depth. The studied also observed a non-significant reduction in femoral flexion relative to anterior pelvis tilt. Although hip angle did not demonstrate a ROM effect due to FAI, limited femur flexion may indicate a functional limit imposed by aberrant morphology (Bagwell et al., 2016). Similarly, Lamontagne et al. (2009b) reported a 9° reduction in pelvic ROM with FAI during the squat. The only difference in sagittal kinetics was a reduction in hip extensor moment reported by Bagwell et al. (2016). The combination of reduced pelvic posterior tilt and diminished hip extensor moment may be indicative of an underlying muscle imbalance. Weakness of the gluteus maximus has been associated with FAI and is known to accompany a more anteriorly (or less posteriorly) tilted pelvis (Casartelli et al., 2011).

In the transverse plane, Lamontagne et al. (2009a) and Kumar et al. (2014) did not observe differences in peak hip internal rotation while Bagwell et al. (2016) found a 6° deficit in the FAI group. Bagwell et al. (2016) had subjects squat to a greater depth than Kumar et al. (2014) which could attribute to the discrepancy. Lamontagne et al. (2009a), on the other hand, used an almost identical protocol and failed to identify a significant difference between FAI and healthy controls. In the Bagwell et al. (2016) study, reduced internal rotation was not accompanied by a change in peak external moment. This combination may be a result of impingement pathomorphology constraining motion as opposed to insufficient muscular control (Audenaert et al., 2012). The only kinetic finding was an increase in peak internal rotation moment at the hip by Kumar et al. (2014). The Kumar et al. (2014) study was limited by a small sample size (7 FAI, 8 control), which were not matched with respect to age or sex.

In the frontal plane, the squat kinematics and kinetic findings are again mixed. Kumar et al. (2014) reported an increase in peak hip internal rotation associated with FAI. No other ROM or kinetic differences were reported across squat studies. Kumar et al. (2014) did not control for foot placement during the squat where the other three studies used consistent placement.

Additional functional testing was examined with stair climbing and drop landings. Stair climbing was examined in one study using preoperative FAI candidates and controls (Rylander et al., 2013). As an ADL, stairs present a unique testing situation that requires a larger hip ROM and more muscular strength than walking (Bergmann et al., 2001). Results indicated a significant 6.2° reduction in hip sagittal ROM, due in large part to a significant 4.8° reduction in hip extension. Maximum internal rotation in the FAI group was significantly diminished by 5°, which is consistent with clinical testing FAI ROM data (Philippon et al., 2007b; Rylander et al., 2013). The combination of decreased internal rotation and extension may act as a pain avoidance strategy in attempting to avoid impingement with the femoral head and acetabulum (Kennedy et al., 2009a; Rylander et al., 2013). Drop landings were studied by Kumar et al. (2014) within the same study used to assess gait and deep squats. The 12-in drop landing height did not illuminate any kinematic or kinetic differences between FAI and healthy controls. Of note, FAI participants landed with their feet positioned more closely together. Drop landings provide a more dynamic task than gait or deep squatting, with larger muscular demands required. The smaller base of support may reduce the muscular strength needed to control the landing (Kumar et al., 2014).

The limited number of three-dimensional movement studies comparing symptomatic FAI with healthy controls have yielded mixed findings. These differences may be attributed to differences in subject samples, small sample sizes, and data collection techniques. Future studies will help identify consistent trends in symptomatic FAI mechanics.

Hip Mechanics

Measured Hip Forces

Direct measurement of hip joint forces is limited due to the invasive methods necessary to gain access to the joint. Telemetering instrumented hip implants allow for direct measurement of these forces *in vivo*. Due to ethical considerations, only those individuals in need of a hip replacement would be potential participants in such research. This imposes potential bias in the findings, as all of the participants would no longer possess their native anatomy and would have undergone a major lower extremity surgery. In the past, individuals in need of hip replacements either represented an older demographic or presented with an anatomic abnormality.

Rydell (1966) was the first researcher to directly measure hip joint forces. In a time before hip replacements were commonplace, existing prostheses were deemed inappropriate for data collection. The study details the daunting process of designing, machining, and instrumenting an implant without the luxury of computer aided design software and advances in electronic circuitry. Only the femoral head was replaced, leaving the native acetabulum in place. In one passage, Rydell describes how, "it is difficult to attach the gauges stably to the moist and slightly greasy surface of the bone" (Rydell, 1966).

Davy et al. (1988) examined one participant with a telemetrized hip implant during several tasks, 31 days after surgery. Forces during gait were 2.6 to 2.8 BW in stance and 2.1 BW in ipsilateral stance. Stair climbing generated a peak force of 2.8 BW. However, data was recorded from one participant quite soon after surgery. As such, it is difficult to generalize the findings to the larger population.

Bergmann et al. (1993) presented force data obtained from two participants with instrumented prostheses in the femoral reference plane. Participants were analyzed while walking and jogging on a treadmill, and stumbling without falling. Only one of the participants could be considered otherwise healthy, as the second participant was chosen due to a neurological condition. The researchers anticipated increased hip forces to be related to the neurological deficit. The healthy participant was found to exhibit 280% to 480% BW when walking 1km/h and 5km/hr respectively. Fast walking and jogging each exhibited 550% BW. Forces were higher in the second, pathological patient.

In 2001, (Bergmann et al.) used a similar research framework with four participants ranging in age from 51-76 years of age. Testing was performed 11-20 months postoperatively and included nine activities. Over ground walking at three speeds, stair climbing and descending, standing up and sitting down, knee bends, and a 2-1-2 footed stance were assessed. Results indicated only small intra-subject variations in hip joint loading during cyclical activities, but much larger inter-subject variations. Walking at 4-km/h was found to generate averge peak hip joint forces of 238% BW. Stairclimbing was found to generate 251% BW climbing up and 260% BW on the way down. These forces were reported to be within range of previous findings (Bergmann et al., 2001; Davy et al., 1988; Rydell, 1966), but lower than Bergmann et al. (1993).

Modeled Hip Forces

Estimation of hip joint forces began long before the age of computers. As early as the 1940s, researchers estimated hip forces during gait to fall somewhere in the vicinity of 4.4 to 4.6 times body weight (Rydell, 1966). The key methodical differences in estimating hip mechanics have been associated with defining how forces are distributed across the hip. Hip forces have traditionally been solved using an inverse dynamics approach with measured kinematics and ground reaction forces, followed by distributing the forces across load bearing structures surrounding the joint (Crowninshield et al., 1978). There have been two approaches to the problem: reduction of the number of factors and optimization. The numerous anatomical structures involved in hip motion present an indeterminate system, with more variables than equations to solve. Paul (1965) and Morrison (1967) used a mechanical simplification strategy. In another approach, Seireg and Arvikar (1973) and Penrod et al. (1974a) proposed the use of a limiting function to constrain the system of equations. Optimization utilizes an objective function to "optimize" and has been extensively used. However, the "best" objective function to use has remained elusive. Static optimization has been criticized for oversimplifying system anatomy and failing to truly encompass muscle dynamics. However, when static optimization was compared with dynamic optimization and the physiological inverse approach, findings demonstrated static optimization was an effective method for hip force estimation in gait (Anderson and Pandy, 2001b; Wesseling et al., 2015).

Hip joint forces have been modeled in numerous studies (Correa et al., 2010; Crowninshield, 1978; Heller et al., 2001; Heller et al., 2005; Koopman and Horsman, 2008; Modenese et al., 2011a; Modenese and Phillips, 2012; Pedersen et al., 1997; Rohrle et al., 1984; Stansfield et al., 2003). Validation of hip forces musculoskeletal models has been approached using *in vivo* measurements and EMG (Lenaerts et al., 2008; Pedersen et al., 1987; Seireg and Arvikar, 1975). Measures of muscle activation through EMG are readily available and relatively noninvasive, but comparison between measured EMG and force output is not consistent or reliable (Anderson and Pandy, 2001b). Similarly, the timing of muscle activations based on EMG data may only roughly approximate modeled muscle activity. Direct comparison of *in vivo* data with simulated model data is important in establishing internal validity (Brand et al., 1994; Heller et al., 2001; Lu et al., 1997; Stansfield et al., 2003). Data obtained by Bergmann et al. (2001), known as Hip98, is open-source for musculoskeletal modeling. This data has been used as a pathway for validation in modeling studies by several researchers (Modenese et al., 2011b; Stansfield et al., 2003; Wesseling et al., 2015). However, care must be taken with the interpretation of comparisons between modeled findings and measured findings. Inter-subject variability has been reported in all direct measurement studies. Thus, a musculoskeletal model that “fits” a particular subject within the Hip98 data may not demonstrate similar results with all subjects. In addition, data is limited by the instrumentation, collection, and possible processing errors incurred by the initial researchers.

Brand et al. (1994) used experimentally collected gait data and direct measurements from a single patient to compare hip joint loading. Using a non-linear optimization, findings indicated measured forces were lower than modeled. The laboratory and implant data were collected several weeks apart, which could account for some force differences. Considering the post-operative time period, there may have been changes in gait mechanics over this time period that resulted in different hip loading characteristics. Loading of the implanted hip may require time to return to a normal force range. Reported peak forces during participant selected gait speed were between 2.5 and 3.5 BW.

Heller et al. (2001) collected femoral loading information via prosthesis in four participants for direct comparison with a custom model. The criterion used was the minimization of muscle force, with a limit on maximum muscle stress. They reported a predominantly compressive force in the femur, with a small shear component. The highest femoral forces were observed in stair climbing, followed by walking (speed not indicated).

Stansfield et al. (2003) made a direct comparison with a musculoskeletal model and the Hip98 dataset. The model utilized a two-stage optimization. First, the optimization minimized maximum muscle stress. Then the sum of muscle and joint forces were minimized (Bean et al., 1988). There was reasonable agreement between modeling with slow, medium, and fast walking speeds, as well as stand to sit tasks.

Modenese et al. (2011b) implemented a new open source lower extremity model referred to as London Lower Limb using anthropometric data from Klein Horsman (2007). The study used kinematic and kinetic data available as part of Hip98 to validate the model with measured force values. Inverse dynamics and static optimization were performed with OpenSim. The objective function was examined at different exponential values from 1 to 15 (Rasmussen et al., 2001), and results indicated hip joint forces increased as the exponential power of the objective function increased. When muscle activation results were overlapped with EMG tracings from previous research (McFadyen and Winter, 1988; Wootten et al., 1990), a quadratic relationship was identified as the most representative of measured data. Increasing the objective function power is associated with increased muscle synergy, while simultaneously allowing a larger number of actuators to contribute to torque production (Dul et al., 1984b; Rasmussen et al., 2001). Unfortunately, with increased muscles acting, moments are created out of the plane the external moment is acting and causes additional actuators to stabilize the hip (Modenese et al., 2011b).

Wesseling et al. (2015) used OpenSim Gait2392 (SimTK, 2016) to compare four different methods of determining muscle forces during gait and sit to stand in five participants. The participants' EMG obtained in the study and the Hip98 dataset were then used for model validation. Two static optimization methods were used. The first was the OpenSim standard static optimization (Delp et al., 2007) with muscle forces resolved by minimizing the square of muscle activations subject to a quadratic optimization criterion (Modenese et al., 2011a; Rasmussen et al., 2001). The second was another static optimization method using a similar cost function with constraints to impact physiological activation (Lenaerts et al., 2008). The third method was CMC, which creates a forward dynamic simulation that utilizes static optimization with feedforward and feedback controls to optimally track position, velocity, and acceleration data that are used to compute muscle excitations and then muscle forces. The fourth approach was a physiological inverse approach (PIA) (De Groote et al., 2009). This method uses a global optimization over the movement cycle with muscle activation squared as the optimization criteria. Results indicated that optimization approach significantly impacts hip joint force, with both static optimization methods producing values more consistent with measured forces. CMC produced the highest values for hip joint force. The second optimization method produced the lowest during gait and sit-to-stand. However, the additional constraints used in the second optimization method did not yield significant differences from the OpenSim standard method. CMC calculated hip joint forces were 238% higher than those reported via direct measurement (Bergmann et al., 2001).

Internal hip joint forces have also been estimated during a maximal hight countermovement jump, jerk, and snatch (Cleather et al., 2013). The hip was modeled using only the right limb following the Klein Horsman (2007) model. Twelve athletic males were examined. Joint forces were normalized to body weight, jump height was not accounted for. Optimization was performed using sum of muscle stress raised to the 20th power (Crowninshield, 1978; Rasmussen et al., 2001). Muscle stress and cross-sectional area were used to calculate muscle forces. Then, internal joint force was estimated by accounting for intersegmental and muscle forces. Mean peak normalized internal hip joint force with standard deviation was reported as 5.5 (1.1) for jumping and 6.0 (3.0) for landing.

Landing Hip Mechanics

The lower extremity mechanics associated with landing from a jump have been thoroughly examined, with a significant focus on the prevention of knee injuries. During a jump landing, a large amount of force is developed while simultaneously incurring large ROM in the hip, knee, and ankle. Subtle variations in landing technique can elicit a wide range of kinematic, kinetic, and energetic outcomes. With recent attention focused on hip pathologies in young athletes, a better understanding of hip mechanics during landing is warranted.

Drop landings are often used to assess landings in a controlled environment that allows for comparisons between participants and groups. A common approach for drop landings uses a platform of known height. The participant begins on top of some type of platform and then steps off to land on one or two force plates below. The participant may land on one foot, both feet, land-and-cut, or land and rebound depending on the study design.

Relationship between landing mechanics and sex. The high prevalence rate of anterior cruciate ligament (ACL) injuries in females has garnered widespread attention from the medical and science communities. Sex differences in landing mechanics during a wide array of sport specific movements have been examined. The preponderance of this research has focused on the knee and specifically how the lower extremities function with respect to the knee. Hip mechanics are typically mentioned, yet seldom fully discussed.

From the ever-growing body of ACL landing mechanics research, several sex differences have emerged in the mechanics of sports actions. Females are reported to demonstrate reduced hip flexion and knee abduction while performing sport actions (Brown et al., 2009; Kernozek et al., 2005; McLean et al., 2005; Weinhandl et al., 2010). During landings, females may also incur larger peak vertical GRF (Decker et al., 2003; Kernozek et al., 2005; Salci et al., 2004) and peak posterior GRF (Kernozek et al., 2005; Weinhandl et al., 2015). In examining the impact with respect joint kinetics, sex differences in the frontal plane indicate females may experience greater hip and knee loading while landing (Brown et al., 2009; Decker et al., 2003; Kernozek et al., 2005; Weinhandl et al., 2010). The differences in GRF, kinematics, and joint kinetics carry over into the energy absorption strategy trends between sexes. Males are more likely to absorb landing energy at the hip, while females are more likely to implement a more knee and ankle dominate energy absorption strategy (Decker et al., 2003; Schmitz and Shultz, 2010; Shultz et al., 2010; Weinhandl et al., 2010). The underlying mechanism for the varying strategy in females may be a result of diminished hip musculature strength (Decker et al., 2003; Shultz et al., 2010).

Differences in frontal (i.e. coronal) plane kinematics and kinetics are the most pronounced findings among sex comparisons of landing (Ford et al., 2006; Garrison et al., 2005; Gehring et al., 2009; Hewett et al., 2004; Hewett et al., 2005; Jenkins et al., 2016; Kernozek et al., 2008; Kernozek et al., 2005; Pappas et al., 2007a; Pappas et al., 2007b; Russell et al., 2006; Weinhandl et al., 2015; Weinhandl et al., 2016). It should be noted, additional findings have failed to identify frontal plane sex differences (Kiriyama et al., 2009; Nagano et al., 2007). Greater knee abduction was reported in females compared to males in unilateral landings from 13.5-cm to 60-cm heights (Ford et al., 2006; Pappas et al., 2007b; Russell et al., 2006). In double leg landings, a greater knee abduction angle was also reported for heights ranging from 40-cm to 60-cm (Gehring et al., 2009; Kernozek et al., 2008; Kernozek et al., 2005; Pappas et al., 2007a; Pappas et al., 2007b). At initial contact during land-and cut maneuvers, females exhibited decreased hip abduction and increased knee abduction. In addition, females also displayed reduced knee abduction ROM (Weinhandl et al., 2016). Knee abduction angle in females may be related to pubescent status, with increased knee abduction angles reported in post-pubertal females (Hewett et al., 2004). Similar findings were reported for pubertal females experiencing a period of accelerated growth (Ford et al., 2010).

Increased knee valgus moment has been associated with an increased risk in ACL injuries (Weinhandl et al., 2015). Internal knee valgus moments are generated by the combined contributions of knee adductor and hip abductor moments. Knee adductor moments have demonstrated a sex effect for bilateral landings (Garrison et al., 2005), while unilateral landings have not displayed the same trend (Weinhandl et al., 2015; Weinhandl et al., 2010). During unilateral landings a greater knee adduction moment was observed in males (Garrison et al., 2005). When examining a land-and-cut maneuver, females were reported to incur increased hip abduction, knee extension and ankle plantarflexion moments.

Differences in the sagittal plane have also been identified (Chappell et al., 2002; Decker et al., 2003; Ford et al., 2003; Hewett et al., 2001; Malinzak et al., 2001); however, the magnitude of these differences is smaller than frontal plane findings (Kernozek et al., 2005; Weinhandl et al., 2015). Compared to males, females landed with a less flexed hip and knee (Salci et al., 2004). Pollard et al. (2010a) reported that females who landed with less hip flexion were more likely to generate greater knee valgus and knee adductor moments. Several studies did not find ROM sex differences with respect to the hip (Kernozek et al., 2005) and knee (Fagenbaum and Darling, 2003). However, females demonstrated greater ankle dorsiflexion (Decker et al., 2003; Kernozek et al., 2005). At initial contact for a land-jump task, females demonstrated less hip and knee flexion compared to males (McLean et al., 2005).

Females have been reported to land with a more erect posture during a double leg drop landing (Decker et al., 2003; Salci et al., 2004). This finding has not been reported in all landing studies and appears to be sensitive to the population, task, and task complexity (Fagenbaum and Darling, 2003; McLean et al., 2005). The knee is the primary mechanism responsible for energy absorption during landing in both sexes. However, females show a propensity to rely on knee and ankle joints for additional energy absorption while males incorporate the hip (Decker et al., 2003; Weinhandl et al., 2015). Montgomery et al. (2014) attempted to equalize task demands between sexes to eliminate the potential bias. Their results found that males still incorporated the hips more for energy absorption, but no sex differences were observed for the knee and ankle. The kinetic energy developed during landing is absorbed in part by eccentric muscle action of the lower extremity extensors (McNitt-Gray, 1993; Schmitz and Shultz, 2010). Thus, in females the knee extensors and plantar flexors are more responsible for energy absorption than the hip extensors, and vice versa for males (Decker et al., 2003). The knee dominant landing strategy observed in female landing mechanics may be attributed in part to diminished hip extensor strength (Pollard et al., 2010a; Zhang et al., 2000).

The relationship between sex and landing mechanics is complex due to the variety of populations studied in an array of landing and sport tasks. The reoccurring themes amongst study findings are that the largest differences are related to frontal plane motion at the hip and knee. These differences are reported in terms of hip and knee positioning at initial contact, maximum angles, and ROMs. Diminished strength in the female hip musculature compared to males in conjunction with altered positioning has been proposed as a risk factor for ACL injuries. The hip dominant approach reported in males may present long term issues not yet fully understood. The implications of using the hip as an energy absorber may have cumulative detrimental impacts of a more insidious nature. Changes in the bony morphology associated with FAI are likely associated with loading and energy absorption characteristics that result in osseous adaptations. A more complete understanding of how the “normal” hip behaves during loading conditions is needed.

Relationship between landing mechanics and limb side. Bilateral differences in landing mechanics have been described in terms of left versus right and dominant versus non-dominant sides. Strength, flexibility, and coordination deficits between sides are associated with an increased injury risk (Hewett et al., 1999; Hewett et al., 1996). With an imbalance, the weaker and/or less coordinated limb may be unable to withstand increased forces or maintain control in an aberrant situation while the stronger limb may experience excessively high loading in an attempt to protect the weaker limb (Weinhandl et al., 2016). The differentiation between dominant and non-dominant limbs has been open to interpretation.

Bilateral differences have been reported in landing research (Brown et al., 2009; Weinhandl et al., 2016). Brown et al. (2009) reported significantly more hip flexion, adduction, and internal rotation, as well as, less knee flexion and adduction at initial contact in the non-dominant leg of a land-and-cut task. The study did not identify any kinetic differences between limb sides. Similarly, Weinhandl et al. (2016) examined land-and-cut maneuvers and reported decreased hip flexion at initial contact on the dominant side. Additionally, a decrease in hip flexion and hip adduction ROM were reported during the landing. Other studies have not indicated significant meaningful differences in bilateral kinematics and kinetics (McLean et al., 2007; Van der Harst et al., 2007b). Limb related data has not yielded clear, consistent findings pertaining to mechanical differences by limb side (Brown et al., 2009; Weinhandl et al., 2016).

In ACL injury research, landing mechanics may relate to an interaction between landing side and sex. An association with the non-dominant limb and increased injury risk has been reported in females (Brophy et al., 2010; Negrete et al., 2007). Within these studies, the non-dominant limb was acting as stabilizer while the dominant leg kicked. Conversely, males were more likely to injure their dominant limb (i.e. kicking limb). Ford et al. (2003) reported increased valgus knee angles in female high school basketball players' dominant limb compared to non-dominant. The same discrepancy was not found in males. Reduced hip abduction and knee adduction at initial contact, as well as, hip abduction and knee adduction ROM were identified in females during unilateral land-and-cut maneuvers (Weinhandl et al., 2016). In addition, increased hip abduction, knee extension, and ankle plantarflexion moments were also reported in females compared to males.

Relationship between landing mechanics and height. As landing height increases, it is reasonable to assume differences in landing mechanics would occur. There were no significant differences identified in vertical or posterior GRF when landing height was increased (Weinhandl et al., 2015). It should be noted, GRF was normalized by body weight and landing height squared. Normalization for landing height allows for the comparison of GRFs while accounting for differences due to landing from varying heights. There was a reduction in knee and ankle energy absorption as landing height increased, but hip energy absorption remained similar. Interestingly, a shift in lower extremity energy absorption with increased landing height did not include a shift toward proximal joints (Montgomery et al., 2014; Weinhandl et al., 2015). Proximal-to-distal energy absorption strategies are believed to expose passive structures to higher forces when landing (Norcross et al., 2010).

Joint moments and work have been reported to increase with increased landing height (Huston et al., 2001; McNitt-Gray, 1993; Yeow et al., 2009, 2010b). The extent of these changes is a function of sex (Huston et al., 2001). When drop landing height was equal to their maximal vertical jumping ability, females displayed less hip abduction at initial contact than males (Weinhandl et al., 2015). Further, females generated higher peak knee extensor and plantar flexor moments during the landing. As landing height increased, females and males modified their mechanics using different strategies. Females exhibited an increased peak knee adductor moment and a decreased peak ankle plantarflexor moment. Conversely, males implemented a reduced peak hip abductor moment (Weinhandl et al., 2015).

Kinetic and kinematic analysis of landing mechanics provides insight into differences associated with sex, landing condition, landing height, and landing limb side. This information describes what forces are being incurred with varying condition and how the motions occurred. However, joint kinetics generated using this approach represent a net value of all the muscles crossing the joint. Musculoskeletal modeling allows for the inclusion of muscle geometry and physiology to estimate joint loading using measured kinetics and kinematics.

Musculoskeletal Modeling

Traditional biomechanical analysis of human movement involves the use of kinematic, kinetic, and EMG data. When used in conjunction, researchers are capable of quantifying segment and angular mechanics on a gross scale. However, these studies are limited in their ability to describe movement dynamics on the muscular level (Delp et al., 2007; Delp et al., 1990a). While the contributions of muscle groups (i.e. moments) and muscle activity (i.e. EMG) can be obtained using traditional methods, individual muscle force is not generally quantifiable. Muscle activity measured using EMG does not directly correlate with segmental motion, due in part to the complex geometry of the musculoskeletal system. Segmental accelerations can be generated by muscles that are not directly attached and joint accelerations applied to joints the muscle does not span (Zajac and Gordon, 1989). Further, the system complexity is compounded by muscles that span multiple joints. A straightforward one muscle to one segment mechanical relationship cannot be used. Movement simulations provide solutions to estimate muscle and joint forces which otherwise are unavailable via noninvasive measurements. These simulations apply multibody movement mechanics within the framework of human neuromuscular anatomy and physiology parameters.

Modeling Steps

Inverse Kinematics. Experimentally collected marker coordinates and scaled model data are synthesized to create generalized model coordinates using inverse kinematics. Using scaled model data, maker trajectories for each time point are fitted with respect to the model, generating generalized coordinates. The resulting "pose" is the mathematical best fit between experimental data and scaled model parameters. Errors incurred during data collection are inherent (e.g. movement artifact), which reduce the likelihood of a perfect fit for the model. Additionally, musculoskeletal models typically do not include the 6 dof for each segment that are included in traditional motion capture. These differences between available model motion and experimental data necessitate the usage of an optimization procedure to match model and experimental motions. Thus, a weighted least squares approach is used to minimize the difference between experimental marker and virtual marker positioning (applied to the model) at each time point:

$$SquaredError = \sum_{i=1}^{marker} w_i\left(\vec{x}_i^{subject} - \vec{x}_i^{model}\right)^2 + \sum_{j=1}^{joint\ angle} w_j\left(\Theta_j^{subject} - \Theta_j^{model}\right)^2$$

The three-dimensional positions of the i^{th} marker for the subject and model are expressed as $\vec{x}_i^{subject}$ and $\vec{x}_i^{model}$ respectively. The three-dimensional positions of the j^{th} joint angle for the subject and model are expressed as $\Theta_j^{subject}$ and Θ_j^{model} respectively. Marker weights, w_i and w_j, can be designated to control how tightly virtual model data tracks the experimental coordinate and joint angle data. Individual marker locations associated with high tracking variability can be given low weighting, while markers needing tight control can be given higher weighting. Segmental cluster data is typically weighted higher than individual data points.

Unfortunately, errors in experimental data collection and processing, as well as, model simplifications and assumptions yield model kinematics that are dynamically inconsistent with measured ground reaction forces and moments. In this case, measured ground reaction force and model marker accelerations violate Newton's second law of motion. Thus, additional non-physiological forces and moments, called residuals, must be added to balance the inconsistency. The following equation represents Newton's second law with the inclusion of residual forces:

$$\vec{F}_{external} = \sum_{i=1}^{segments} m_i \vec{a}_i - \vec{F}_{residual}$$

where $\vec{F}_{external}$ is measured GRF minus the body weight vector, $\vec{a}_i$ and m_i are the translational COM and mass of the *i*th segment, and $\vec{F}_{residual}$ is the residual force. Residual moment is calculated similarly using GRF moment. Conceptually, residual forces and moments would approach zero in a perfect modeling experiment. However, when experimental data is considered, residuals represent non-negligible quantities. In order to maintain dynamic consistency, residuals must be applied to a body within the model.

OpenSim utilizes an algorithm known as the Residual Reduction Algorithm (RRA) to manipulate marker trajectories and model mass parameters to minimize residuals while maintaining dynamic consistency. When residual forces are low, the model's movement can be attributed to the internal joint torques with more certainty. The RRA represents a form of dynamic forward simulation, where the model begins at an initial time point with kinematics derived during inverse kinematics and steps forward incrementally to the pre-defined end time point, calculating force values for all model actuators to drive the inverse kinematics. The skeletal structure and joint anatomy described previously are assigned torque actuators for each dof. The torque actuators generate internal torques necessary for motion. The six degrees of freedom between the pelvis and the ground are each modeled with a residual actuator, where translations are represented as forces and rotations represented as torques.

Estimating Muscle Forces. Static and dynamic optimization solutions can be used to estimate muscle forces. Static optimization uses an inverse dynamic while dynamic optimization is based on forward dynamics. In both types of optimization, segmental motion is estimated based on internal joint torques and applied forces.

Static Optimization. The concept of static optimization began in the early 1970s as a method of calculating lower extremity muscle forces in different static positions (Seireg and Arvikar, 1973). Within a few years, variations of the first static optimization would follow using the wrist (Penrod et al., 1974b) and walking (Seireg and Arvikar, 1975). Numerous static optimization studies have followed (An et al., 1989; An et al., 1984,Crowninshield, 1978 #217; Crowninshield and Brand, 1981a; Crowninshield and Brand, 1981b; Dul et al., 1984a; Dul et al., 1984b; Hardt, 1978; Herzog and Leonard, 1991; Kaufman et al., 1991a, b; Penrod et al., 1974b; Prilutsky et al., 1997), yet the underlying goal of determining the set of muscle forces necessary to recreate a desired action while minimizing a measurable quantity remains consistent. The theory is that in a system where multiple muscle force solutions are possible, the use of a particular combination of muscle forces must be based on some prescribed criterion (Crowninshield, 1978). Advances in computational power have enabled the development of increasingly complex model and optimizations. An et al. (1989) stated, “the problem has been to develop a muscle model that is phenomenologically correct without being overwhelmingly complex for practical applications.”

The net joint moments at each time step can be determined using static optimization to calculate individual muscle forces. The inputs needed for static optimization are the joint torques calculated during inverse dynamics (possibly RRA) and the scaled model with lines of muscle action. Given this information, a linear or non-linear algorithm can be applied to determine a solution that generates muscle forces while minimizing the performance criterion and reproducing joint torques. The measurable quantity to be minimized is referred to as the performance criterion. Cost functions and objective functions can similarly be used to constrain the system dependent on the research or movement objective.

A variety of performance criterion have been suggested to best represent human movement. Initial studies focused on minimizing muscular effort or force (Seireg and Arvikar, 1973). However, this criterion did not account for muscle geometry nor physiology. In some instances, small muscles were tasked to generate unreasonable amounts of force. The physiological cross sectional area (PCSA) of a muscle was then included to determine muscle stress, using constraint equations to limit PCSA (Crowninshield, 1978). This new criterion satisfied the need for geometrical consideration. In a slightly different approach, Pedotti et al. (1978) used a non-linear performance criterion by using the sum of muscle stress squared, while also accounting for the force length velocity muscle properties. This approach was argued to incorporate geometry and physiology, as muscle stress squared is a measure of fatigue (Crowninshield and Brand, 1981a). This concept was further extended with the concept of minimizing global stress (An et al., 1989). By minimizing global stress, load sharing increased between synergistic muscles. The global minimization of muscle activation squared was then proposed (Kaufman et al., 1991a, b).

With Static Optimization, the cost function is evaluated at each time step independently, where frames prior to and after do not directly dictate muscle forces in the given frame. This characteristic of static optimization can present issues when adjacent frames result in dramatically different muscle forces. Muscle physiology does not support an on-off function of muscle mechanics, which indicates the muscle forces are not indicative of true forces. However, muscle forces from the previous frame can be used as inputs from the subsequent frame's calculations. The cost function is still evaluated to determine muscle forces at the given frame, but they are less likely to incur dramatic on-off muscle actions. Cost functions are sensitive to initial conditions, thus starting near a solution increases the likelihood of similar behavior. A reduction in computational time is an additional benefit. Once static optimization is complete for each frame, the simulated muscle forces can be used to reproduce model motion.

Dynamic Optimization. A muscle driven dynamic simulation provides a complement to traditional movement data and provides the opportunity to examine cause and effect relationships between muscular excitations and resulting motions (Delp et al., 2007). A muscle driven dynamic simulation of movement is built upon a dynamical model of the musculoskeletal system (including environmental interaction) and muscle excitation inputs. A musculoskeletal model for dynamic simulations requires specific parameters describing skeletal structure and joint kinematics, muscle paths and activations, as well as excitation-contraction coupling relationships (Delp et al., 1990b). Muscle excitations serve as model inputs and can be calculated by solving the optimization problem at each time point (Neptune and Hull, 1998). A working simulation can provide muscle and joint force data previously unavailable to researchers. Dynamical simulations can be evaluated based upon a task criterion or level of agreement between measured kinematics, kinetics, and EMG and simulated outcomes. Unfortunately, identifying a combination of muscle excitations that reproduce the desired coordinated multibody movement comes at a high computational cost. OpenSim (SimTK, 2016) is an open source simulation environment which allows dynamic modeling of human motion. The implementation of multibody simulations specifically within OpenSim will be discussed.

Dynamic optimization has several benefits over static optimization, which are purported to better incorporate muscle physiology. In a dynamic optimization, the entire motion sequence is used when minimizing performance criterion. This yields a continuous flow of muscle forces that are time dependent. Further, the number of muscles contributing to torque production are not limited by degrees of freedom as can be found with linear optimization techniques (e.g. linear algebra). The resulting muscle forces are less likely to be impacted by errors incurred in data collection and processing as the entire process is analyzed. However, these advantages come with a tremendous computational cost (Anderson and Pandy, 2001b).

In OpenSim, a modified form of dynamic optimization called computer muscular control (CMC) utilizes static optimization in conjunction with feedforward and feedback controls to drive the model towards the inverse kinematic solutions. A static optimization criterion is used to distribute forces across synergistic muscles with an overall objective to minimize global error between measured and simulated kinematics. The differential equations within are referred to as state equations and are used to describe activation dynamics, musculotendon dynamics, and Newton's second law of motion. A forward dynamic model of human movement begins with a dynamical model of the musculoskeletal system including model interaction with the environment and necessary muscle excitation inputs.

Model Components

Skeletal Model. The skeletal model serves as the primary framework for developing a modeling system. Boney segments are created as rigid bodies, based upon polygons fitted to the surface of specific bone (Delp et al., 1990a). The base model for these bones are stored in individual files and can be scaled based on a participant's anthropometric data. Each bone file, classified as a body, contains the mass properties and associations with other visible objects. The relationship between bones is prescribed as a joint, with each joint containing a parent and child segment. Joint motion is defined in terms of coordinate and kinematic transforms with respect to the parent segment.

The motion of a joint is a relationship between two bodies (parent and child). The body represents a moving frame of reference with a prescribed COM and inertial properties, called the body frame. A joint frame of reference, the joint frame, can be defined with respect to the body frame. The rotational and translational transforms of the child with reference to the parent define joint motion. Specific joint options include weld, pin, slider, ball, ellipsoid, free, and custom joints. Pin and ball joints are commonly used in the lower extremities. A pin joint utilizes one coordinate about a common x-axis, while a ball joints allow three rotational coordinates about the x, y, and z axes.

The generalized model created above can be scaled to an individual's anthropometry for improved subject specificity. Individualized model scaling is achieved by fitting a calibration model with experimental maker data. Segment positioning in the general reference frame must be defined by a least three non-collinear markers which provide angels and translations in six dof. The fitting of a calibration model results in appropriate scaling of all the musculoskeletal element geometries and mass properties.

Environmental Interaction. Interaction with the ground can be modeled using springs and dampers in series (Anderson and Pandy, 1999; Anderson and Pandy, 2001a). Three dimensional springs allow simultaneous force in the vertical, fore-aft, and transverse directions. Vertical force varies exponentially with respect to vertical foot positioning. Fore-aft and transverse forces vary linearly with position. As a drawback to this method, the damped spring will always apply forces to the model. With exponential decay, this force becomes negligible at a designated height above the ground. Expressing model-ground compliance with the damped-spring method minimizes vertical displacement below ground level and saves computational cost (Anderson and Pandy, 1999).

Muscle models. The skeletal model is actuated by muscles, represented as specialized force elements. Muscle actuator files define muscle properties and geometry. Specific actuator properties dictate activation and contraction dynamics, fiber length, and geometry. Muscle actuators are connected to bones as discrete points with origins and insertions consistent with their anatomical structure (Pandy and Anderson, 2000). In special cases where muscles have broad origins or insertions, the muscle can be modelled using multiple muscles to better simulate effective force actions and moment arms. Muscle geometry is defined by the path associated with muscle action. One method of determining muscle geometry is with straight-line muscle paths and via points (Delp et al., 1990a; Hoy et al., 1990; Jensen and Davy, 1975). A centroid-line method of determining geometry has also been proposed, which defines muscle path through the locus of cross-sectional muscle centroids (Jensen and Davy, 1975). This approach may better represent muscle action; however, the ability to repeatedly recreate centroid location across multiple subjects and positions is questionable.

The fit of straight-line muscle paths can be amended for non-linear muscle actions through via points (Anderson and Pandy, 1999). Via points attach to bone, similar to the origin and insertion. The muscle would then run from the origin, through via points and into the insertion. An indefinite number of via points can be applied to sub-divide a muscle line into separate smaller linear paths. While each segment is linear, the summative effect of multiple segments can create a more curvilinear path. Further, via points can be activated and deactivated as a function of joint position. The resulting application of muscle force provides a better approximation of muscle wrapping around bones and joints.

Musculotendon Actuation. Musculotendon mechanical behavior can be modeled as a three element Hill type muscle in series with tendon (Zajac and Gottlieb, 1989). The force-length-velocity relationship for muscles and the elastic properties of tendon are integrated using a lumped parameter model. A contractile element characterizes force length and force velocity characteristics of muscle, while series and parallel elastic elements represent passive and active stiffnesses, respectively (Zajac and Gordon, 1989; Zajac and Gottlieb, 1989).

Muscle force can be calculated using five muscle metrics: peak isometric muscle force, optimal muscle fiber length, pennation angle, tendon slack length, and maximum shortening velocity. First, peak isometric muscle force represents the largest amount of force a given muscle can produce without changing length. The measure is scaled with respect to PCSA to ensure small muscle forces are not tasked with supra-physiological forces (Brand et al., 1986; Delp et al., 1990b; Wickiewicz et al., 1983). Initially presented by Haxton (1944), PCSA describes overall muscle architecture and includes measures of muscle mass, pennation angle, fiber length and density (Wickiewicz et al., 1983). The second and third metrics, optimal muscle fiber length and pennation angle, are derived from muscle conditions when peak isometric force is exerted. Both measures are measured from cadaveric data where optimal fiber length is the measure of fiber length at peak isometric force and pennation angle is the orientation of the fibers in relation to axis of muscle shortening. The fourth metric, tendon slack length, is a measure of the visible tendon plus the aponeurosis. Inclusion of the aponeurosis makes it difficult to obtain direct measurements without compromising muscle function. When the tendon is displaced beyond slack length (Zajac and Gottlieb, 1989), elastic force is generated following a linear force curve

(Pandy et al., 1990). Using a linear force curve simplifies the amount of strain energy stored in the tendon; however, the resulting effect on actuator performance is relatively small. The non-linear toe region of the force displacement graph occurs where forces are low. Then, as forces increase a linear approximation provides a good fit. Subtle changes in tendon slack length have a significant impact with respect to the joint angle where peak force occurs and the resulting magnitude of the force (Hoy et al., 1990; Zajac and Gottlieb, 1989). The ratio of optimal muscle-fiber length to tendon slack length is a commonly used metric in predicting muscle coordination in simulations (Delp et al., 2007). The last metric, maximum shortening velocity, is commonly assumed as a constant and independent of the specific muscle at $10s^{-1}$ (Pandy et al., 1990). The muscle specific maximum shortening velocity can be represented as $10\left(\frac{optimal\ muscle\ fiber\ length}{s}\right)$. The constant value for shortening velocity represents the summative result of slow, intermediate, and fast twitch muscle fiber actions (Zajac and Gottlieb, 1989).

Musculotendon actuators are defined for each muscle acting within the model. The time rate of change for muscle force, for any given muscle, can be represented as:

$$\dot{F}^{MT} = f(F^{MT}, l^{MT}, v^{MT}, a^{m}); where\ 0 < a^{m} < 1$$

Hence, $\dot{F}^{MT}$ is a function of musculotendon force (F^{MT}), musculotendon length (l^{MT}), musculotendon shortening velocity (v^{MT}), and activation (a^{m}). The resultant muscle force is representative of active and passive components and related to a specific musculotendon length, velocity, and activation (Delp and Loan, 1995).

Excitation-Contraction Coupling. Muscle activation is mediated by calcium concentrations in response to motor unit action potentials (Ebashi and Endo, 1968). The correlation between calcium dynamics and muscle excitation generates a continuum between activated and relaxed muscle states, in which the muscle does not instantly become activated or deactivated and a delay exists between excitation and activation (Zajac and Gottlieb, 1989). A first order process using muscle activation and excitation can be used to model the delay.

$$\dot{a^m} = (u - a^m) \cdot \left[\frac{u}{\tau_{rise}} + \frac{(u+1)}{\tau_{fall}}\right]; \text{ when } u > a^m$$

$$\dot{a^m} = \frac{(u-a)}{\tau_{fall}}; \text{ when } u < a^m$$

In this representation, the time rate of change in muscle activation is only dependent on activation and excitation at a given time point. Excitation (u) is a measure of net neural drive and varies from fully deactivated (0) to fully activated (1). Previous studies have included two variables for excitation to capture recruitment and stimulation (Happee, 1994; Pandy and Zajac, 1991; Raasch et al., 1997). In multi-joint simulations, incorporation of these quantities did not impact model outcome to the extent of time constant values (Zajac and Gottlieb, 1989). The time constants (τ) vary with respect to activation and deactivation (Raasch et al., 1997). When excitation is less than activation the muscle is relaxing, thus the time constant for contraction does not impact activation. Time constants for contraction vary from 12-20ms, while relaxation constants vary from 24-200ms (Zajac and Gottlieb, 1989).

Estimation of Muscle Excitations. As discussed previously, the goal of a dynamic simulation, such as CMC, is to identify the set of muscle excitations capable of reproducing desired kinematics. The process begins by using the experimentally collected kinematic data (with or without RRA) to identify desired accelerations. The accelerations of the generalized coordinates can be found using

$$\ddot{\vec{q}} = \overleftrightarrow{A}^{-1}(\vec{q}) \cdot \left\{\vec{G}(\vec{q}) + \vec{C}\left(\vec{q}, \dot{\vec{q}}\right) + \overleftrightarrow{R}(\vec{q}) \cdot f_m + \vec{E}\left(\vec{q}, \dot{\vec{q}}\right)\right\}$$

where $\vec{q}$, $\dot{\vec{q}}$, $\ddot{\vec{q}}$ represent the generalized coordinates, velocities, and accelerations, $\overleftrightarrow{A}^{-1}$is the inverse of the system mass matrix, $\overleftrightarrow{R}$ is the muscle moment arm matrix, f_m is the vector of muscle forces, and $\vec{G}$, $\vec{C}$, and $\vec{E}$ are the vectors of generalized forces associated with gravity, Coriolis and centripetal forces, and environmental interactions respectively.

Once desired accelerations are identified for the current time point, the acceleration, velocity, and position data are fed forwards to determine accelerations a short time later (e.g. 0.01s later). Feedback gains are employed to modify the calculated acceleration to best fit the velocity and position data. Static optimization is then used to obtain muscle activations that ultimately recreate the calculated accelerations and inverse dynamics data. Using force length velocity properties and contraction dynamics, muscle forces can be approximated using activation. With the activations now known, the necessary muscle excitations for the future time point are solved for and implemented at the current time point, which allows for the incorporation of excitation-activation dynamics. In the final step, simulated excitations are fed into the forward dynamic model. At each time step, the state equations are integrated to output acceleration, activation, and contraction dynamics. The simulation progresses until the predetermined end state is reached. Simulated and experimental joint angles and ground reaction forces are compared using an RMS approach.

OpenSim Gait2392

Several models of human motion are available within OpenSim. OpenSim is an open source, collaborative effort which allows the creation and visualization musculoskeletal movement while generating meaningful mechanical and physiological data outputs (Delp et al., 2007). The program is an attempt to streamline musculoskeletal modeling and encourage simulation repeatability across institutions (Seth et al., 2011). However, generalized comparisons of simulation data must be avoided due to the dependence on experimental data and the high vulnerability to measurement error due to collection equipment, methodologies, and pre-processing (Reinbolt et al., 2011).

Gait2392 is a three-dimensional, 23-dof model of the musculoskeletal system designed to analyze lower extremity mechanics within OpenSim. Darryl Thelen, Ajay Seth, Frank C. Anderson, and Scott L. Delp served as the primary architects. However, the model represents collaborative research efforts in defining bone, muscle, and joint geometry as well as inertial properties and actuator characteristics (Anderson and Pandy, 1999; Delp et al., 1990b; Hoy et al., 1990; Inman, 1976; McConville et al., 1980; Nisell et al., 1986; Spector et al., 1980; Stredney, 1982; Wickiewicz et al., 1983; Yamaguchi and Zajac, 1989). It is important the underlying equations, assumptions, simplifications, and sources of error within a model be understood when designing research studies and interpreting results.

Kinematics

The kinematic definitions described within the following sections pertain to the unaltered Gait2392 model; specific parameters can be refined as warranted based on study design and research questions. The unscaled model in Gait2392 is 1.8 m tall with a body mass of 75.16 kg. Ten rigid segments (one pelvis, four segments for each limb) representing the lower extremity are set in motion by 92 actuators, representing 76 different muscles.

Bone geometry. The lower extremity model is comprised of the pelvis, femur, patella, tibia, talus, calcaneus, and toes. Each bone has been recreated in three-dimensions to represent a scalable, generalized boney structure. Pelvis and thigh bones were reconstructed using polygon mesh to recreate slices within a plane. The recreated structure was then digitized for three-dimensional coordinates (Delp et al., 1990b). The tibia, talus, calcaneus, and toe structures were defined using higher order computer fitting technique (Stredney, 1982, 1985).

Muscle geometry. Muscles are modeled as muscle actuators. Each muscle actuator has a muscle path that dictates the direction of applied forces. The origin and insertion of each muscle are based on anatomical landmarks (Delp et al., 1990b). Muscles with broad insertions are divided into multiple muscle actuators that share an origin and have insertions that are in line with the true muscle. Muscle paths are modeled as close to anatomically correct as possible, which requires additional constraints beyond the direct path from origin to insertion. Special points called via points are utilized along the muscle path to account for bone wrapping with position changes and retinaculum movement constraints. Some lower extremity muscles are not included in the model due to difficulties in recreating realistic muscle paths and moment arms. A complete listing of the muscles included is provided in Appendix B.

Segment geometry. The lower body rigid body segments in Gait2392 are the torso, pelvis, thigh, shank, foot, and toes. The foot is comprised of the talus and the calcaneus. Reference frames are fixed in each segment. The orientation of each segment's reference frame is defined below (Delp et al., 1990b).

1. The pelvic frame is located at the midpoint of the line connecting anterior superior iliac spines.
2. The femoral frame is located at the center of the femoral head.
3. The tibial frame is located at the midpoint between medial and lateral femoral condyles.
4. The talar frame is located at the midpoint between apices of the medial and lateral malleoli.

5. The calcaneal frame is located at the most interior, lateral point on the posterior surface of the calcaneus.
6. The toe frame is located at the base of the fifth metatarsal.

Joint geometry. Joints connect adjacent segments and are defined as the hip, knee, ankle, subtalar, and metatarsophalangeal joints. Joint motions are prescribed as relative motion between segments. For each joint, parent and child segments are defined. Translations and rotations are defined for the child segment in reference to the parent.

Each joint is defined with motions representative of joint mechanics. The hip joint is straightforward and can be modeled as a ball and socket joint. The knee joint is a more challenging system, comprised of the femur, patella, and tibia. Yamaguchi and Zajac (1989) simplified knee motion to one dof which accounts for tibiofemoral kinematics while still allowing the mechanical advantage generated by the patella (Nisell et al., 1986). The transformations between reference planes of the three bones can then be expressed as a function of knee angle (Delp et al., 1990b). The ankle, subtalar, and metatarsophalangeal joints are modeled as frictionless revolute joints (Delp et al., 1990b). Orientation and location of joint axes for the ankle was described by Inman (1976). In order to maintain articulation, the metatarsophalangeal joint is oriented 8° on a right-handed axis.

Dynamics

Movement of the musculoskeletal model described above is directly related to the models inertial properties and the muscle actuators that generate motion. Inertial properties of a segment quantify that segments resistance to change in state of motion, with respect to mass density and velocity. These properties shape how each segment will move when toque is applied.

Inertial properties. Mass distribution characteristics for five participants were reported by Anderson and Pandy (1999) using the detailed methodology of McConville et al. (1980). The segmental mass and inertial properties used in Gait2392 were based on this data (scaled by 1.05626), with the exception of the hindfeet and toes. For these segments, a size ten shoe was used to estimate the total segmental volume. Assuming density was constant, segment masses were estimated as density integrated over volume. Segmental lengths were obtained from (Delp et al., 1990b). The inertial parameters for each segment are provided in Appendix C.

Muscle actuators. The musculoskeletal geometry and parameters of 43 actuators in each leg were taken from Delp et al. (1990b) with an additional six actuators to account for pelvic-HAT kinematics from Anderson and Pandy (2001a). A scaling factor has been applied to the actuators taken from Delp et al. (1990b) to account for differences in modeled strength and experimentally measured joint-torque angle relationships. Muscle strengths within the model are derived from both Delp et al. (1990b) and Anderson and Pandy (2001a), depending on which data more accurately represented physiologically measured values. Musculotendon properties are quantified comparably to Hoy et al. (1990). PCSA was determined while incorporating findings from Spector et al. (1980), Friederich and Brand (1990), and Wickiewicz et al. (1983). Muscle optimal fiber length and pennation angle were taken from Wickiewicz et al. (1983) when possible. The fiber length was scaled as a ratio of sarcomere length at peak muscle force (2.8-µm) versus measured sarcomere length (2.2-µm) (Wickiewicz et al., 1983). For unavailable muscles, muscle optimal fiber length and pennation angle were taken in the anatomical position from Friederich and Brand (1990).

Performing simulations utilizing a well-developed, yet not overly complex, model such as Gait2392 allow for decreased computation time while providing a thorough examination of hip joint forces. The kinematic and kinetic data needed for model inputs can be collected non-invasively, with limited time commitment, as well as, relatively low financial cost. The combined methodologies allow for the collection and computation of hip joint forces from a large participant sample given limited resources.

CHAPTER 3: METHODS

The overall objective for this dissertation was to examine hip mechanics in asymptomatic individuals during landing maneuvers. To this end, there were three separate aims: (1) identify bilateral and sex differences in HJF, GRF, and lower extremity kinematics during a drop landing; (2) determine if landing task and sex influence HJF, GRF, and lower extremity kinematics; and (3) examine the relationships between drop landing height and sex with HJF, GRF, and lower extremity kinematics.

Participants

For this study, a total of 83 participants were recruited from a medium sized university and the surrounding communities. Inclusion criterion stated all participants were between 18 and 30 years of age and physically active. Physically active was defined as performing at least 30 minutes of exercise 3 times a week. At least one of these activity sessions was required to consist of repetitive jumping and landing, such as basketball or volleyball. Participants from both sexes and all ethnicities were included. The screening process required each subject to complete a general medical history form. Scoring on these questionnaires helped identify members of each group and describe our study sample. Individuals who reported a lower extremity injury in the past six months, any history of lower extremity surgery, or a health condition which may affect their ability to land from a jump were excluded from participation. Participants were free of pain on testing days.

Instrumentation and Equipment

Testing was performed in the Neuromechanics Lab (ODU, Norfolk, VA, USA). Spandex shorts and lab standard tennis shoes (Air Max Glide, Nike, Beaverton, OR, USA) were worn for testing sessions. Three-dimensional (3-D) marker coordinate data was collected at 200-Hz using an eight-camera Vicon MX motion analysis system (Vicon, Centennial, CO, USA). Ground reaction forces (GRF) were measured synchronously at 2000-Hz using two flush mounted force plates (Bertec, Columbus, OH, USA). 3-D lower extremity kinematics and kinetics were calculated through Visual3D (v4.95, C-Motion Inc., Rockville, MD). OpenSim (v3.1, http://simtk.org) was used to create a 3-D model for landing and land-and-cut simulations (Delp et al., 2007). Further data reduction and batch processing was performed using MatLab (MathWorks, Natick, MA, USA). Statistical analyses were performed using custom MATLAB code and SPSS (SPSS Inc., Chicago, IL, USA).

Experimental Protocol

Upon arrival for data collection, all participants received a full explanation of the study purpose and procedures. Prior to data collection, participants were informed of study procedures and provided the written informed consent in accordance with university institutional review board policies. Once consent was given, a health history questionnaire was completed. Participants were asked to change into snug fitting spandex shorts and lab standard tennis shoes. Height and weight were documented.

Prior to data collection, the motion capture system and force plates were calibrated according to manufacturer’s specifications. The force plates were manually zeroed. The motion capture system was calibrated using a standard reference wand with retro-reflective markers in precise locations. To define the data collection volume, the wand was waved in 3-D throughout the desired movement space. Each marker was identified within the program and tracked for the duration of the system calibration trial.

Single reflective markers were placed on the skin over specific anatomical landmarks (Weinhandl et al., 2010) with adhesive tape for calibration purposes only. Calibration-only markers were placed bilaterally on the acromioclavicular joints, iliac crests, greater trochanters, medial and lateral knee epicondyles, medial and lateral malleoli, first metatarsal heads, and fifth metatarsal heads. Marker tracking in movement trials utilized eight cluster plates positioned on the upper torso and pelvis, as well as, bilateral thighs, shanks, and feet. For these tracking clusters, four retro-reflective markers were attached to semi-rigid, molded Orthoplast (Johnson & Johnson, Raynham, MA, USA). Thigh and shank clusters were secured using a cohesive wrap and McDavid groin wraps (McDavid, Woodridge, IL, USA). Foot clusters were affixed to lab shoes using industrial strength Velcro. Pelvis and trunk clusters were secured with custom neoprene straps.

When all markers were placed on the participant, a three-second static trial was collected. For this trial, the subject was asked to stand motionless with arms crossed high over the chest and each foot on a separate force plate. Calibration-only markers were removed and technical markers remained.

Participants completed five successful unilateral landings and land-and-cut maneuvers on their right and left limbs from three heights. All participants completed the landing-only task and 42 participants completed landing-only and land-and-cut tasks. Testing heights for drop landings were 30-cm, 40-cm, and 50-cm. Land-and-cut tasks were performed from 40-cm. Selected heights are absolute and commonly used in drop landing research (Bruton et al., 2013). The landing task required participants to begin on top of a box with prescribed height standing on the contralateral limb desired for landing. The intended landing limb was lifted with hip and knee flexed approximately 90°. When ready, the participant stepped from the platform and landed on the force plate. The land-and-cut task required participants to drop from a prescribed height, land on the designated limb, and immediately cut to the contralateral side along a 1-m wide path oriented at 45°. The 1-m wide path ensured participants maintained a cutting angle of 40°–50°. For all landing tasks, participants were instructed to keep their arms folded across their chest throughout the maneuver. A successful landing trial was defined as participants' ability to (1) perform the task without stepping down or jumping up from the box, (2) land with their entire foot on the force plate, (3) land without hopping or jumping (4) land without the contralateral limb contacting the ground, and (5) maintain crossed arms. A successful land-and-cut trial was defined as participants' ability to (1) perform the task without stepping down or jumping up from the box, (2) land with their entire foot on the force plate, (3) execute the side cut without hesitation while staying within the boundaries, and (4) maintaining crossed arms. No instructions were provided on landing technique or style. Landing height order and limb side were counterbalanced across participants. Participants were allowed to perform practice trials before each condition. Data collection for each participant lasted approximately 30 minutes.

Data Analysis

Data reduction was implemented through Visual3D (v4.95, C-Motion Inc., Rockville, MD). Raw 3-D marker coordinate and GRF data were low-pass filtered using a 4th-order, Butterworth filter with a cutoff frequency of 10-Hz (Kristianslund et al., 2012). The landing for all tasks were defined as initial contact (IC) with the force plate through maximal knee flexion (MF). IC was identified as the first-time point GRF surpasses 20-N. MF was defined as the time point when knee flexion reaches the first global minimum (i.e. the minimum of knee extension). Data was time normalized to 101 data points to define 100% of the landing phase. Comparisons were made at each percent of motion.

Using Visual3D, a kinematic model comprised of eight skeletal segments (trunk, pelvis, and bilateral thighs, shanks, and feet) was created from the standing calibration trial (Weinhandl et al., 2010). An unweighted least squares approach was used (Spoor and Veldpaus, 1980). The hip joint centers were defined as 25% of the distance from the ipsilateral to the contralateral greater trochanter markers (Bennett et al., 2016; Weinhandl et al., 2010). Joint centers for the knees and ankles were defined as the midpoints between medial and lateral epicondyles (Grood and Suntay, 1983), as well as, medial and lateral malleoli markers (Wu et al., 2002), respectively. 3-D bilateral ankle, knee, and hip angles were calculated using a joint coordinate system approach (Grood and Suntay, 1983). 3-D joint kinetics were calculated using a Newton-Euler approach (Bresler and Frankel, 1950) and reported in the distal segment reference frame. Dempster (1955) was used for body segment parameter data.

OpenSim v3.3 (SimTK, 2016) was used to simulate landing from IC to MF. For this study, a 19-dof, 8 segment generic musculoskeletal model was scaled for each subject using individual anthropometric data from the calibration trial (Delp et al., 1990b). For simplicity, the head, arms, and torso were combined into a rigid body segment referred to as HAT. The remaining seven segments were the pelvis, bilateral thighs, shanks, and feet. The pelvis was the principal segment with position and orientation identified by six generalized coordinates. Remaining segments branched from the pelvis in an open-chain system.

Pelvic translation and rotations were modeled with 6-dof. Lumbar and hip motions were modeled as 3-dof ball and socket joints (Anderson and Pandy, 1999), while knee and ankle motions were modeled as a one dof revolute joints (Delp et al., 1990b). Tibiofemoral translation was constrained as a function of knee flexion (Nisell et al., 1986; Yamaguchi and Zajac, 1989). Ninety-two Hill-type contractile elements in series with tendon were used to actuate the model; 43 for each leg and 6 for the torso (Anderson and Pandy, 1999; Delp et al., 1990b; Zajac and Gordon, 1989).

The inverse kinematics problem was solved at each frame to compute model joint angles. This generated model coordinate values to match experimental data using a least squares approach while accounting for constraint weights (Delp et al., 2007). Inverse dynamics were performed using calculated joint angles and measured GRF data. The results provided an estimation of forces and moments that caused the motion to occur. The net reaction forces and moments for each joint which maintain equilibrium were determined (Kuo, 1998).

Muscle forces were estimated using static optimization at each time step using inverse dynamics outputs, external load data, scaled model and optimization criteria. Output forces and moments were filtered at 6-Hz. Residual actuators were added to the top most segment, the pelvis, for 3-D dynamic inconsistencies in forces and moments. Optimization criteria was the sum of muscle stress squared (Anderson and Pandy, 2001b; Crowninshield and Brand, 1981a; Kaufman et al., 1991a). RMS reserve joint torques were verified to be below 5% of net joint moments (Hicks et al., 2015). Sample trials were compared with published modeling data in landings (Hicks et al., 2015).

Hip joint forces (HJF) were calculated using the JointReaction analysis algorithm within OpenSim (SimTK, 2016). HJF represents the intersegmental joint reaction forces between adjacent segments, acting through the center of the joint. The internal loads represent the summative result of all loads applied to the body. HJF were reported as the pelvis on the femur in the femoral reference frame and acting through the joint center (Steele et al., 2012).

Discrete kinetic variables were peak GRF (pGRF), peak HJF (pHJF), and time of pHJF. Resultant values of GRF and HJF were calculated using their 3-D components and normalized by body weight. GRF and HJF discussed refer to resultant measures except when explicitly stated. pGRF and pHJF were identified as the first global maximum for each respective measure.

Time to pGRF and pHJF were determined from IC. Average loading rates for pGRF and pHJF were calculated as the respective maximal force value divided by time to peak force. Discrete kinematic variables were 3-D hip, knee flexion, and ankle dorsi/plantar-flexion measures at IC and pGRF. ROMs were calculated for 3-D hip, knee flexion, and ankle dorsi/plantar-flexion for the impact phase (ROM_I) and landing phase (ROM_L). HJF, GRF, and 3-D hip, knee flexion, and ankle dorsi/plantar-flexion angles were exported as continuous data for each trial. Subject averages were computed at each percentage of landing phase. Subject averages were averaged again to generate group means for sex and condition (e.g. limb side) at each percentage of the landing phase.

Statistical Analysis

Statistical analyses were performed using custom MATLAB code (MathWorks, Natick, MA, USA) and SPSS (SPSS Inc., Chicago, IL, USA). For all aims, participant demographics are presented with mean and standard deviation for weight and height. All discrete variables were checked for outliers, normality, and homogeneity prior to statistical analyses. Effect sizes were quantified for each study using eta squared (η^2) and calculated as the sum of squares for each factor divided by the total sum of squares for each factor, interaction, and error terms.

Aim 1

The first aim of this dissertation was to examine sex and bilateral differences in hip mechanics during a drop landing. Differences between sex and limb side for HJF, GRF, and 3-D hip, knee flexion, and ankle dorsi/plantar-flexion measures were assessed using mean ensemble curve analysis with corresponding two standard deviation intervals for each percentage of the landing. Statistically different areas of the landing phase were identified where the two standard deviation intervals of the mean ensemble curves between sexes did not overlap. Differences in pHJF, pGRF, timing of pGRF and pHJF, loading rate of pHJF and pGRF, as well as, 3-D hip, knee, and ankle ROM_I and ROM_L between sexes and limb sides were assessed via separate 2x2 (sex x limb side) ANOVAs ($p<0.05$). When significant differences were present in ROM, the maximum and minimum joint angles for the designated measure were analyzed using two-tailed t-tests. Paired t-tests were used for limb comparisons while unpaired t-tests were used for sex comparisons.

Aim 2

The second aim was to investigate the effects of sex and a 40-cm drop landing maneuver (landing-only versus land-and-cut) on hip mechanics. If significant bilateral differences were reported in *Aim 1,* limb side would have been examined. Since no bilateral differences were indicated only sex and landing tasks were included as independent variables. Differences between sex and landing maneuver HJF, GRF, as well as, 3-D hip, knee flexion, and ankle dorsi/plantar-flexion kinematics were assessed using mean ensemble curve analysis with corresponding two standard deviation intervals for each percentage of the landing. Statistically different areas of the landing phase were identified where the two standard deviation intervals of the mean ensemble curves between landing maneuvers or sexes did not overlap. Differences in pHJF, pGRF, timing of pGRF and pHJF, loading rate of pHJF and pGRF, 3-D hip, knee flexion, and ankle dorsi/plantar-flexion measures at IC, as well as, ROM_I and ROM_L, between sexes and landing tasks were assessed via separate 2 x 2 (sex x landing tasks) ANOVAs ($p<0.05$).

Aim 3

The third aim was to examine the effects of sex and landing height (30-cm versus 50-cm) on hip mechanics during landing. If significant bilateral differences were reported in *Aim 1,* limb side would have been examined. Since no bilateral differences were indicated only sex and landing height were included as independent variables. Differences between heights and sexes for HJF, GRF, and 3-D hip, knee flexion, and ankle dorsi/plantar-flexion kinematics were assessed using mean ensemble curve analysis with corresponding two standard deviation intervals for each percentage of the landing. Statistically different areas of the landing phase were identified where the two standard deviation intervals of the mean ensemble curves between landing heights did not overlap. Differences in pHJF, pGRF, timing of pGRF and pHJF, loading rate of pHJF and pGRF, 3-D hip, knee flexion, and ankle dorsi/plantar-flexion measures at IC, as well as, ROM_L and ROM_I, between sexes and conditions were assessed via separate 2 x 2 (sex x landing height) ANOVAs ($p<0.05$).

CHAPTER 4: HIP MECHANICS DURING UNILATERAL DROP LANDINGS

Introduction

Groin injuries have traditionally encompassed a broad spectrum of injuries to the hip and groin area (Griffin et al., 2016; Weir, 2013). Over the past decade, increasing awareness of hip specific conditions, such as femoroacetabular impingement (FAI), has facilitated enhanced diagnostic tools and highlighted a paucity in our understanding of young adult hip mechanics (Orchard, 2015). FAI is the pathomechanical process created by anatomical abnormalities of the proximal femur and/or the acetabulum causing mechanical abutment within the hip (Ganz et al., 2003; Ito et al., 2001; Leunig et al., 2009; Packer and Safran, 2015). Increasing evidentiary trends indicate FAI may be a causal factor in the development of secondary osteoarthritis in young, active adults (Ganz et al., 2003; Tannast et al., 2007b). As a result, FAI is postulated to be a significant and growing cause of disability in otherwise healthy active adolescents and young adults (Ganz et al., 2003). In a cohort study of FAI patients younger than 50-yrs, hip arthroscopy treatment costs were approximately $23,000 compared to over $99,000 for non-operative treatments. In patients receiving surgical treatment, a 2.3 increase in quality of life years was also reported over a 10-year period. FAI may present a greater economic disability than both shoulder arthroscopy and ACL tears (Mather et al., 2016).

Increased forces between the femur and acetabulum are commonly proposed as a mechanism for damage in traditional hip osteoarthritis; however, it is unclear if FAI mechanics are related to increased hip force magnitudes, directions, or areas of force application (Ganz et al., 2003; Ito et al., 2001). In cadaveric studies, incremental landing loads applied to knee joints yielded tibiofemoral cartilage deformation and degeneration (Yeow et al., 2009). Knee and hip OA present with similar biomechanical pathogeneses ultimately due to an inability of joint structures to maintain healthy tissue (Brandt et al., 2008). The directions, magnitudes, and frequencies of these forces determine the mechanobiology of the articular cartilage (Carter et al., 2004) and result in cartilage maintenance, growth, or degeneration. The mechanical joint response is further impacted by soft cartilaginous viscoelasticity and interstitial fluid pressure (Li et al., 2005). Given the viscoelastic properties of bone and cartilage, loading rates incurred during landings are an important descriptor of the overall joint loading.

During a jump landing, a large amount of force is developed while simultaneously incurring large ranges of motion (ROM) in the hip, knee, and ankle (Weinhandl et al., 2015). Subtle variations in landing technique can elicit a wide range of kinematic and kinetic outcomes. Sex and bilateral differences in landing mechanics have yielded varied results that are strongly related to the specific task employed (Brown et al., 2009; Weinhandl et al., 2015). With respect to sex differences, the reoccurring themes amongst study findings are related to frontal plane motion at the hip and knee positioning at initial contact, maximum knee flexion, and ROM. Females have been proposed to land with a stiffer landing configuration compared to males (Decker et al., 2003). In addition, females have been associated with significantly higher vertical ground reaction forces (GRF) during landings (Pappas et al., 2007a; Van der Harst et al., 2007b). Inter-limb investigations have not yielded clear, consistent findings pertaining to mechanical differences by limb side (Brown et al., 2009; Weinhandl et al., 2016), potentially indicating sex differences in landing are bilateral.

With recent attention focused on hip pathologies in young athletes, a better understanding of hip mechanics during landing is warranted. Unfortunately, hip contact forces cannot be directly measured via non-invasive measures. These forces can, however, be estimated through musculoskeletal modeling using measured kinematics and kinetics (Delp et al., 2007). These methods have been successfully used to estimate hip contact forces in normal and pathologic populations (Cleather et al., 2013; Modenese et al., 2011b). A comprehensive understanding of hip joint loading during landing in a asymptomatic population would provide a foundation to begin examining pathological hips. The purpose of this study was to explore what, if any, relationships exist between sex and limb side hip mechanics during unilateral drop landings.

Methods

Eighty-three asymptomatic, physically active participants between the ages of 18 and 30 volunteered for study participation. Upon arrival, all participants received a full explanation of study purpose and procedures. Prior to data collection, participants provided written informed consent in accordance with university institutional review board policies. Participants were recruited from a medium sized university and the surrounding communities. Physically active was defined as performing at least 30 minutes of exercise three or more times a week. At least one of these activity sessions consisted of repetitive jumping and landing, such as in basketball or volleyball. Participants completed a general medical history questionnaire that was used to identify members of each group and describe the study sample. Individuals who reported a lower extremity injury in the past six months, any history of lower extremity surgery, or a health condition which may have affected their ability to land from a jump were excluded from participation. Spandex shorts and lab standard tennis shoes (Air Max Glide, Nike, Beaverton, OR, USA) were provided for testing sessions.

Three-dimensional (3-D) marker coordinate data were collected at 200-Hz using an eight-camera Vicon MX motion analysis system (Vicon, Centennial, CO, USA). GRFs were measured synchronously at 2000-Hz using two flush mounted force plates (Bertec, Columbus, OH, USA). Single reflective markers were placed bilaterally on the acromioclavicular joints, iliac crests, greater trochanters, medial and lateral knee epicondyles, medial and lateral malleoli, first metatarsal heads, and fifth metatarsal heads. (Weinhandl et al., 2010) Marker tracking in movement trials utilized rigid cluster plates with four retro-reflective markers positioned on the upper torso and pelvis, as well as, bilateral thighs, shanks, and feet. A static calibration trial was collected with all markers in place. For individual participant calibration, the participant stood with arms crossed high over the chest and each foot on a separate force plate. Calibration-only markers were then removed, leaving the cluster markers in place.

Participants then completed three successful unilateral drop landings on their right and left limbs from 40-cm. The landing task required participants to begin on top of a 40-cm box standing on the contralateral limb desired for landing (i.e. standing on the left leg for a right leg landing). In the starting position, the intended landing limb was raised with hip and knees flexed approximately 90°. When ready, the participant stepped from the platform and performed a one-footed landing on the force plate. Participants were instructed to keep their arms folded across the chest throughout the landing. A successful landing trial was defined as participants' ability to (1) perform the task without stepping down or jumping up from the box, (2) land with their entire foot on the force plate, (3) land without hopping or jumping, (4) land without the contralateral limb contacting the ground, and (5) maintain crossed arms. No additional instructions were provided on landing technique or style. Landing limb side order was counterbalanced. Practice trials were allowed before data collection.

Data reduction was implemented through Visual3D (v4.95, C-Motion Inc., Rockville, MD). The landing phase was defined as initial contact (IC) with the force plate through maximal knee flexion (MF). IC was identified as the first-time point vertical GRF exceeded 20N. MF was defined as the time point when knee flexion reached a global maximum. Continuous data was time normalized to 100% of the landing phase for comparison at each percent of motion from IC to MF. The impact phase was defined from IC through the first peak in resultant GRF. Raw 3-D marker coordinate and GRF data were low-pass filtered using a 4th-order, Butterworth filter with a cutoff frequency of 10-Hz (Kristianslund et al., 2012).

Using Visual3D, a kinematic model comprised of eight skeletal segments (trunk, pelvis, and bilateral thighs, shanks, and feet) was created from the standing calibration trial (Weinhandl et al., 2010) using an unweighted least squares approach (Spoor and Veldpaus, 1980). Hip joint centers were defined as 25% of the distance from ipsilateral to contralateral greater trochanter markers (Bennett et al., 2016; Weinhandl et al., 2010). Joint centers for the knees and ankles were defined as the midpoints between medial and lateral epicondyles (Grood and Suntay, 1983) and malleoli markers (Wu et al., 2002), respectively. Visual3D was used to perform inverse kinematics on marker trajectories and anthropometrics to fit the Gait2392 model. Gait2392 is a model composed of 8-segments, 19 degrees of freedom (dof), and 92 Hill-type muscle actuators (Delp et al., 1990a). The eight segments and respective dof are: head-arms-trunk segment (3 rotations), pelvis (3 translations and 3 rotations), the hips (3 rotations), and the knees and ankles (1 rotation each).

The inverse kinematics problem was solved at each frame to compute model joint angles. This generated model coordinate values that closely matched experimental data using a least squares approach while accounting for constraint weights (Delp et al., 2007). Inverse dynamics analysis was performed using calculated joint angles and measured GRF data. The net reaction forces and moments for each joint to maintain equilibrium were determined (Kuo, 1998). Muscle forces were estimated using static optimization at each time step using inverse dynamics outputs, external load data, scaled model and optimization criteria. Output forces and moments were filtered at 10-Hz (Kristianslund et al., 2012). Residual actuators were added to the principal segment, the pelvis, for 3-D dynamic inconsistencies in forces and moments. The sum of muscle stress squared was used for optimization criteria (Anderson and Pandy, 2001b; Crowninshield and Brand, 1981a; Kaufman et al., 1991a). RMS reserve joint torques were verified to be below 5% of net joint moments (Hicks et al., 2015). The hip joint forces (HJF) were calculated using the joint reaction analysis algorithm within OpenSim. The HJFs represent the intersegmental joint reaction forces between femur and pelvis segments, acting through the center of the hip joint. The HJFs are expressed as the pelvis on the femur in the femoral reference frame (Steele et al., 2012). Timing of peak force represents time from IC. Due to the nature of the hip joint, resultant forces were utilized to capture overall hip mechanics. Resultant GRF and HJF were calculated from force components and normalized by body weight. All GRF and HJF discussed henceforth represent resultant measures except when specified. Loading rates for HJF and GRF were calculated as the respective peak resultant force measure divided by the time from IC to peak resultant force measure.

Statistical analyses were performed using custom MATLAB code (MathWorks, Natick, MA, USA) and SPSS (SPSS Inc., Chicago, IL, USA). Discrete kinematic variables are 3-D hip, knee flexion, and ankle dorsi/plantar-flexion ROM during the impact (ROM_I) and landing (ROM_L) phases. Discrete kinetic variables are peak GRF (pGRF), pGRF timing, peak HJF (pHJF), pHJF timing, GRF loading rate, and HJF loading rate. The HJFs, GRFs, and 3-D hip angles were exported as continuous data for each trial and averaged for each subject and limb side. All discrete variables were checked for normality and homogeneity prior to statistical analyses. Separate 2x2 (limb x sex) ANOVAs were used to assess differences between each discrete variable with respect to sex and limb side ($p<0.05$). Effect size was quantified using eta squared (η^2) and calculated as the sum of squares for each factor divided by the total sum of squares for sex, limb side, sex-limb side interaction, and error terms. When significant differences were present in ROM, the maximum and minimum joint angles for the designated joint angle were analyzed using unpaired t-tests ($p<0.05$). Differences between sexes for continuous HJF, GRF, and 3-D hip kinematics were assessed using mean ensemble curve analysis with corresponding two standard deviation intervals for each percentage of the landing. Statistically different areas of the landing phase were identified where the two standard deviation intervals of the mean ensemble curves between sexes and sides did not overlap.

Results

Three participants' data was not included in analysis. One participant was excluded due to errors in data collection. The last two female participants to undergo data collection were excluded to maintain balanced sex groups. Forty females (height = 1.64 ± 0.06-m, mass = 61.19 ± 8.66-kg) and 40 males (height = 1.79 ± 0.08-m, mass = 79.78 ± 13.53-kg) were used for data analysis. Five trials were collected for each condition, however only the first three successful trials were used for analysis.

Group ensemble curves for GRF and HJF are provided in Figures 1 and 2. Group mean ensemble curves for joint angles are provided in Figures 3 and 4. There were no significant differences in continuous GRF, HJF, 3-D hip, knee flexion, or ankle dorsi/plantar-flexion measures between sexes or limb sides at any percentage of normalized landing phase. The effects of sex and limb on the kinetic and kinematic discrete variables are provided in Table 1. Across all variables examined, there were no significant effects for limb side or interaction effects between sex and limb.

Significant main effects for sex were identified for pGRF, ROM_I hip flexion, ROM_I hip adduction, ROM_I knee angle, ROM_L hip adduction, and ROM_L hip rotation. No other significant main effects were reported for sex. Females performed unilateral landings with 0.17 BW higher pGRF than males ($F_{1,156} = 6.650$, $p = 0.011$, $\eta^2 = 0.041$). For ROM_I, females displayed 2.0° greater hip flexion ROM ($F_{1,156} = 12.289$, $p = 0.001$, $\eta^2 = 0.073$), 1.5° greater hip adduction ROM ($F_{1,156} = 12.428$, $p < 0.001$, $\eta^2 = 0.111$), and 2.7° more knee flexion ROM ($F_{1,156} = 17.897$, $p < 0.001$, $\eta^2 = 0.100$). Females utilized 4.3° more hip adduction ROM_L ($F_{1,156} = 29.455$, $p < 0.001$, $\eta^2 = 0.159$) and 1.8° less hip rotation ROM_L ($F_{1,156} = 6.534$, $p = 0.012$, $\eta^2 = 0.040$).

Independent t-tests were used to examine sex differences in pooled limb data for maximum and minimum joint angles where ROMs indicated main sex effects. For ROM_I, males incurred greater maximum hip adduction (-15.6°± 7.3° versus -12.8°± 7.2°) and knee flexion (-15.5°± 5.4° versus -13.7°± 4.6°), and reduced maximum hip flexion (14.9°±13.9° versus 20.1°±12.5°) compared to females ($p < 0.050$). When examining ROM_L, males incurred greater maximum hip adduction (-6.7°±6.8° versus -1.2°±7.8°) and minimum hip rotation (-7.8°± 6.7° versus -5.4°± 5.5°) compared to females ($p < 0.050$).

Table 1.
Group means ± STD for study one discrete variables.

	Males	**Females**	**Right**	**Left**
Peak Force (BW)				
pGRF#	3.47 ± 0.40	3.64 ± 0.41	3.57 ± 0.42	3.54 ± 0.40
pHJF#	9.36 ± 2.18	9.86 ± 1.61	9.57 ± 1.74	9.64 ± 2.11
Loading Rate (BW/s)				
pGRF	58.85 ± 11.17	61.44 ± 10.85	61.39 ± 11.57	58.90 ± 10.43
pHJF	136.26 ± 65.09	135.95 ± 55.69	133.58 ± 61.50	134.22 ± 0.80
Time to Peak Force (ms)				
pGRF	60.42 ± 7.90	60.60 ± 7.63	59.71 ± 8.41	61.31 ± 6.97
pHJF	66.60 ± 30.38	77.29 ± 40.19	70.65 ± 33.19	73.25 ± 38.62
ROM_I (°)				
Hip Flexion#	6.80 ± 2.99	8.79 ± 4.03	7.86 ± 3.88	7.73 ± 3.48
Hip Adduction#	4.51 ± 1.88	5.96 ± 2.24	5.28 ± 2.26	5.19 ± 2.13
Hip Rotation	6.27 ± 3.14	5.67 ± 2.80	5.96 ± 2.93	5.98 ± 3.05
Knee Flexion#	20.68 ± 3.94	23.33 ± 4.06	21.48 ± 4.18	22.53 ± 4.20
Ankle Dorsiflexion	33.02 ± 9.14	34.17 ± 8.20	32.55 ± 9.07	34.64 ± 8.17
ROM_L (°)				
Hip Flexion	16.52 ± 8.28	19.15 ± 9.61	18.07 ± 9.17	17.61 ± 8.95
Hip Adduction#	13.39 ± 4.63	17.68 ± 5.29	15.68 ± 5.68	15.39 ± 5.14
Hip Rotation#	10.67 ± 5.02	8.90 ± 3.56	9.68 ± 4.57	9.90 ± 4.30
Knee Flexion	48.12 ± 9.86	49.05 ± 8.14	48.81 ± 8.85	48.36 ± 9.24
Ankle Dorsiflexion	44.55 ± 10.45	45.51 ± 7.81	44.23 ± 9.72	45.83 ± 8.65

Significant main effect for sex ($p < 0.05$)

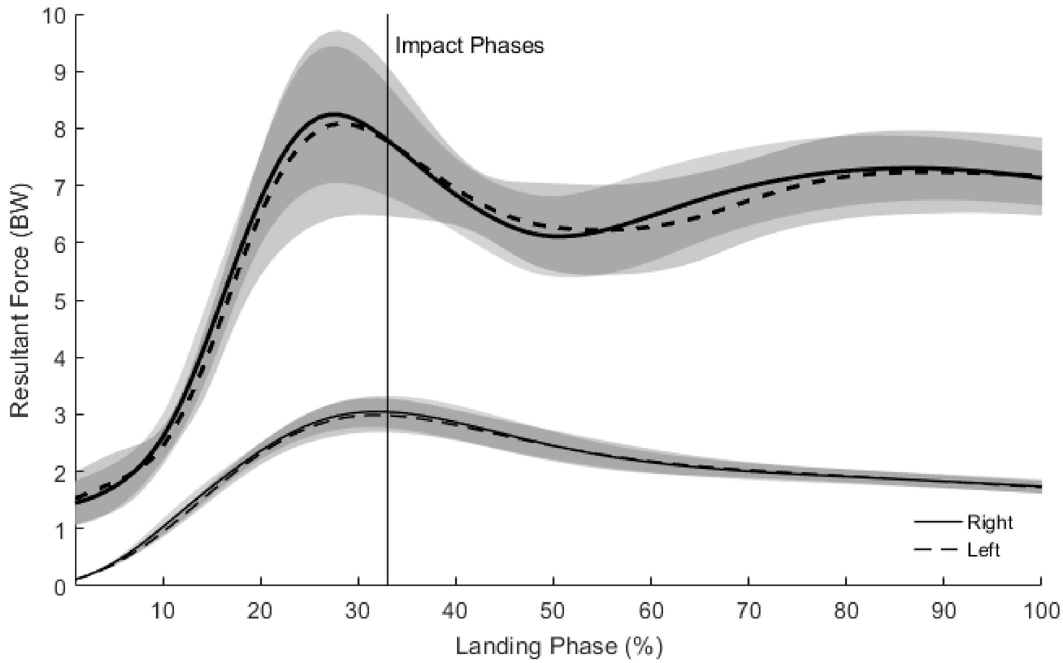

Figure 1. Study one continuous normalized group mean GRF and HJF (bold) ± two STD for limb side.

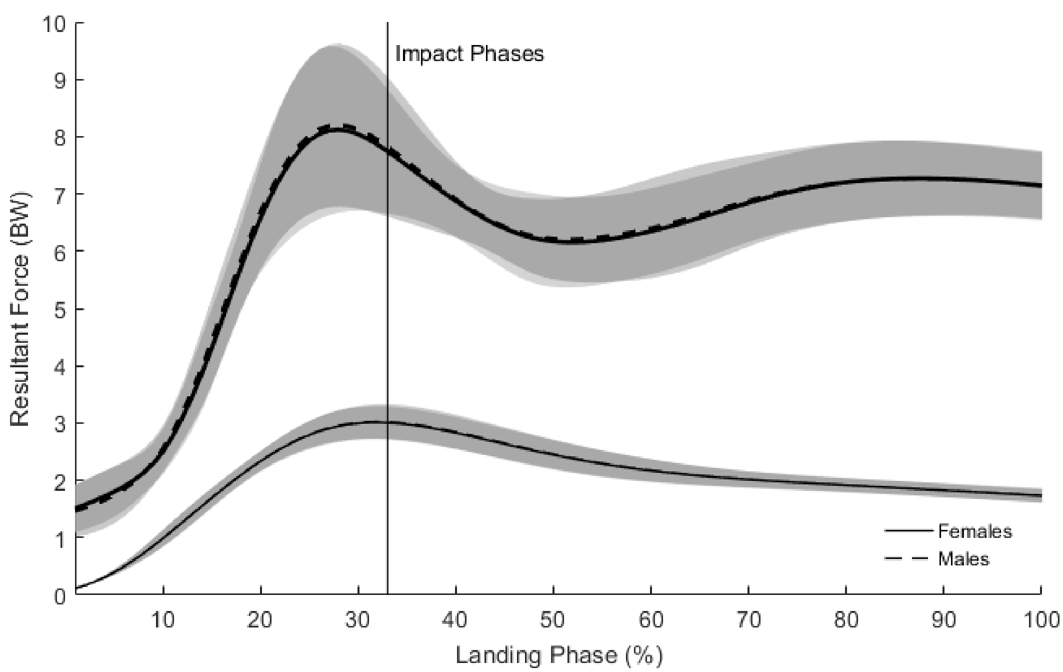

Figure 2. Study one continuous normalized group mean GRF and HJF (bold) ± two STD for sex.

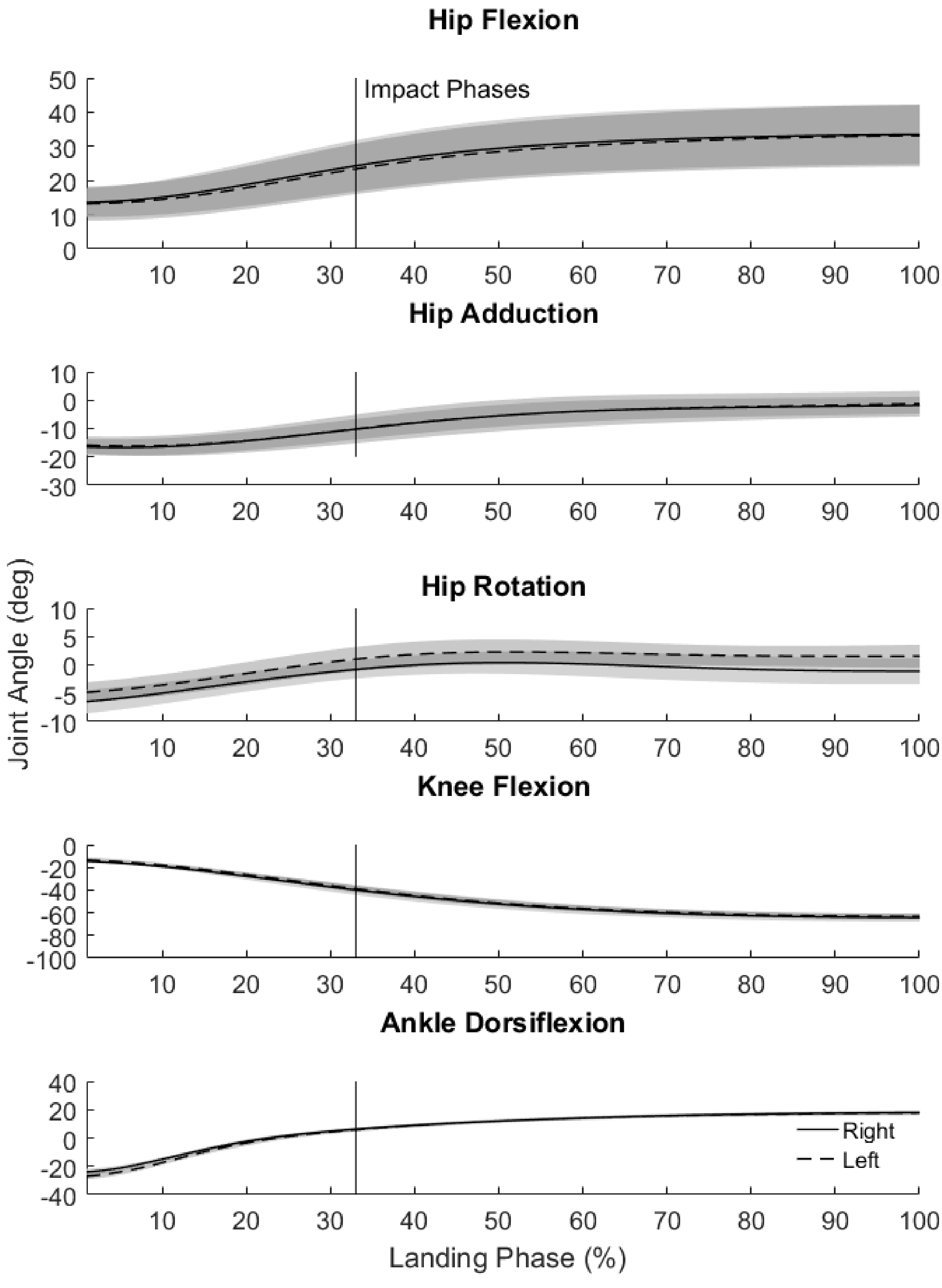

Figure 3. Study one continuous normalized group mean joint angles ± two STD for limb side.

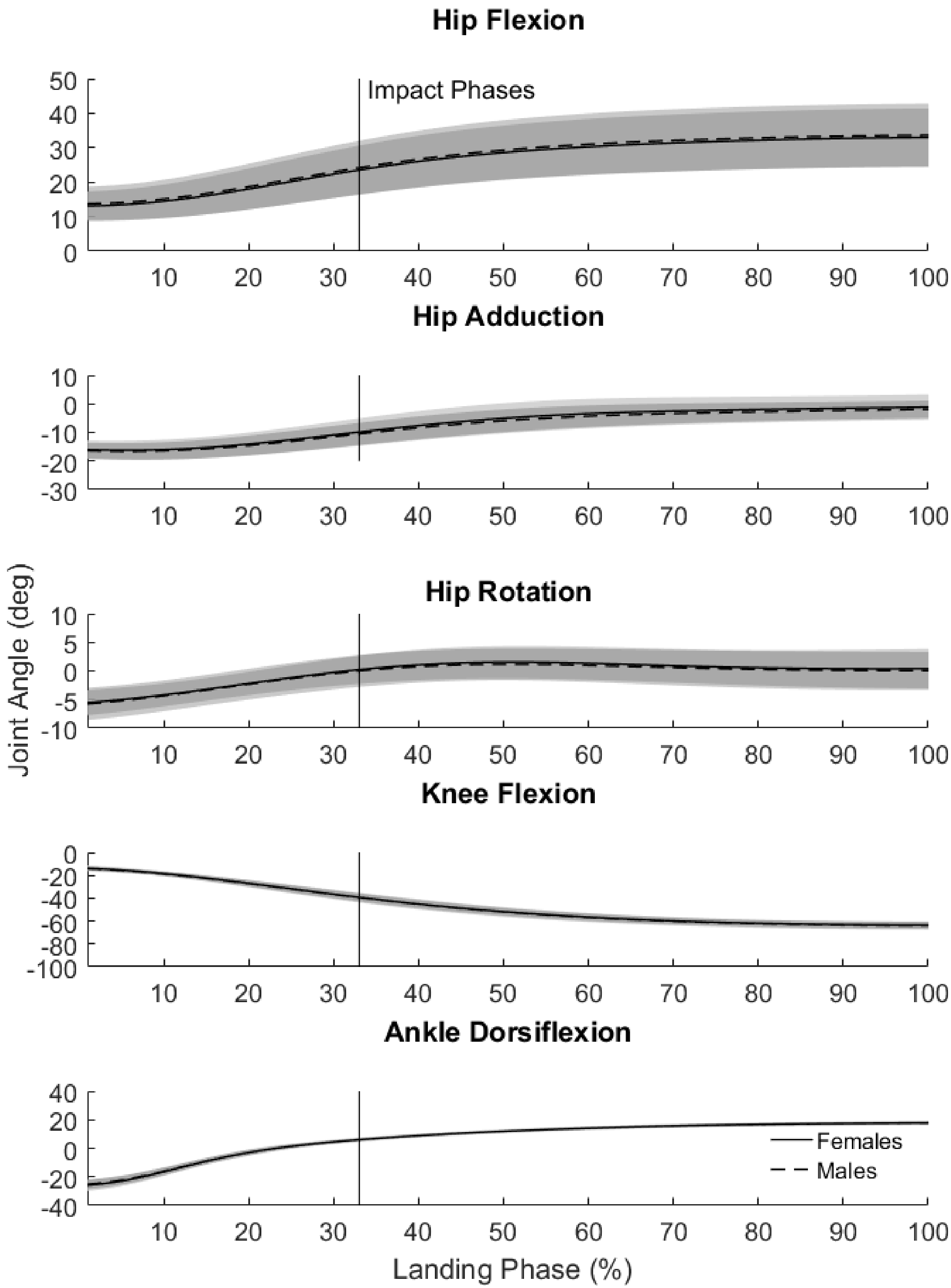

Figure 4. Study one continuous normalized group mean joint angles ± two STD for sex.

Discussion

The number of patients diagnosed with hip pathologies has risen significantly in the past decade (Griffin et al., 2016). Before we can further investigate hip pathomechanics, an understanding of hip mechanics in asymptomatic, active, young adults is needed. The exploratory nature of this study was designed to broadly examine the relationships associated with landing mechanics, specifically at the hip. Using a relatively large-scale modeling sample size, sex and bilateral differences in GRF, HJF, peak force loading rate, peak force timing, and lower extremity kinematics during unilateral drop landings were assessed. Examination of continuous resultant GRF, resultant HJF, 3-D hip, knee flexion, and ankle plantar/dorsi-flexion measures did not reveal significant differences over a normalized landing phase. GRF tracings of the landing phase demonstrated a similar unimodal force time curve reported in vertical GRF landing studies (Schmitz et al., 2007). Similar HJF loading characteristics are evident during the impact phase, however the HJF curve presents a slight secondary peak between the end of the impact phase and peak knee flexion. Limited studies have assessed hip forces during dynamic activities, such as the single leg landing task used in the current study. The HJF measures reported in this study are higher than those observed in *in vivo* studies (Bergmann et al., 2001; Bergmann et al., 1993) and comparable with landing studies *in silico* given the drop height utilized (Cleather et al., 2013).

Examination of the discrete measures teased apart differences corresponding to specific time points and ranges. Our results demonstrated females experience greater resultant pGRF during unilateral landing, without a corresponding sex difference in resultant pHJF. The significantly higher pGRF reported in female landings is consistent with previous findings for vertical and posterior GRFs (Decker et al., 2003; Kernozek et al., 2005; Pappas et al., 2007a;

Salci et al., 2004; Schmitz et al., 2007; Weinhandl et al., 2015). Loading rates and timing from IC to pGRF and pHJF also failed to demonstrate sex effects.

ROM was assessed during impact and landing phases to examine kinematic differences between IC and pGRF and MF respectively. During the impact phase, high GRFs must be safely dissipated by the hip, knee, and ankle. Females generated higher pGRF forces, however, they also experienced larger hip and knee sagittal ROMs during the impact phase. The increased pGRF in females may have been due to landing with a more extended knee at IC. In addition, the increased use of sagittal ROM observed in females may have contributed to the absence of GRF loading rate sex differences. Frontal plane hip measures remained negative throughout impact and landing phases, thus yielding only hip abduction. During both phases, females incorporated larger hip frontal plane ROM, specifically due to lower hip abduction measures (e.g. maximum hip adduction). In similar unilateral landings, Pappas et al. (2007a) found a similar range of hip adduction motion, but did not identify sex differences in hip kinematics. However, positioning of the contralateral limb during landing was not specified and may have contributed to pelvis positioning. Hewett et al. (2006) reported similar trends of decreased hip abduction in female unilateral landings without sex differences in sagittal ROM. Insufficient strength of the hip muscular has been proposed as a key component of the alterations in hip adduction and is associated with an increased risk of lower extremity injury (Hewett et al., 2006; Padua et al., 2005). Interestingly, the only increased ROM observed in males was landing phase hip rotation and associated with greater hip external rotation. All other landing and impact phase ROMs did not indicate sex differences.

Bilateral differences were not reported for any of the kinetic or kinematic variables analyzed, suggesting both limbs landed with similar mechanics. Clear trends in limb differences have yet to be established. A similar lack of bilateral differences has been reported in unilateral hopping (Van der Harst et al., 2007b) and varied heights of unilateral landing (McPherson et al., 2016). However, it should be noted these studies utilized a dominant versus nondominant limb designation. For the exploratory nature of the current study, right and left side designations were implemented. The sample population used in the current study were asymptomatic, young adults which were generally active in sports requiring jumping and not primarily recruited from sports associated with task specific limb differences (i.e. soccer or tennis). As such, significant limb dominance or muscle imbalances were not considered significant intrinsic factors. When considering emerging trends in pathological hips, FAI is estimated to occur bilaterally in up to 75% of radiographic diagnoses which may submit both hips experience similar forces (Allen et al., 2009; Ganz et al., 2008; Klingenstein et al., 2013).

Interpretation of results and findings should be tempered by several methodological limitations. First, the absolute drop height of 40-cm may have represented different levels of task demand based on individual participant jump heights (Weinhandl et al., 2015). Second, HJF were estimated based on simplified musculoskeletal modeling and subject to sources of error associated with marker placement, movement artifact, and muscle force estimates (Hicks et al., 2015). Third, for model simplicity, knee motion was modeled as a 1-dof hinge joint which may have masked sex differences in knee adduction. Lastly, the study sample represents a generalized asymptomatic population and the presence of abnormal hip anatomy is unknown.

Moving forward, inclusion of joint energetics and muscle activations would greatly enhance overall understanding sex differences in hip mechanics during landings. Sex differences in peak GRF and joint kinematics have been associated with energy absorption strategies. Males are suggested to be more likely to absorb landing energy at the hip, while females are more likely to implement a more knee and ankle dominate energy absorption strategy (Decker et al., 2003; Schmitz and Shultz, 2010; Shultz et al., 2010; Weinhandl et al., 2010). Further exploration of these factors may help illuminate the underlying mechanisms responsible for the relationship between GRF and HJF loading characteristics.

CHAPTER 5: THE EFFECTS OF SEX AND LANDING TASK ON HIP MECHANICS

Introduction

Research on femoroacetabular impingement (FAI) has increased in recent years, yet the etiology is not well understood (Diamond et al., 2014; Packer and Safran, 2015; Reiman and Thorborg, 2015). Etiological theories have proposed genetic factors, prior hip morphology (e.g. slipped capital femoral epiphyses), and athletic participation as potential factors (de Silva et al., 2016; Packer and Safran, 2015). In particular, an association between high-level sports participation in adolescence and impingement morphology is rapidly gaining support (Agricola et al., 2012; Kapron et al., 2011; Siebenrock et al., 2011). Prevalence of FAI in adolescent and young adults varies to some degree by sport, with an increased incidence in cutting sports compared to flexibility, contact, impingement, or endurance based sports (Nawabi et al., 2014). In separate sport specific studies, athletes participating at high intensity levels of football, ice hockey, basketball, and soccer, beginning in childhood were at an increased risk of developing impingement morphology (Agricola et al., 2012; Gerhardt et al., 2012; Siebenrock et al., 2011). The combination of repetitive hip loading with flexion and internal rotation during cutting sports may present excessive physiological demands during sensitive adolescent skeletal development (Clohisy et al., 2013; de Silva et al., 2016; Siebenrock et al., 2011). Early exposure to aggressive sporting activity in adolescence may expose the proximal femoral epiphysis to excessive stress and may contribute to development of boney pathomorphologies (Beck et al., 2004; Byrd and Jones, 2011).

The relationships between landing mechanics during sport actions and potential injury risk have received a great deal of attention. Unilateral landings have been proposed to present a different task demand than bilateral landings, with unilateral landings associated with more ‘at risk' mechanics (Boden et al., 2000). Landing and cutting are two commonly performed sports actions attributed to an increased injury risk, both of which are often performed unilaterally. Although similar in nature, specific kinematic and kinetic differences have been associated with drop jump landings and cutting tasks (O'Connor et al., 2009). During these motions, combinations of knee abduction movements and excessive loading have been associated with knee injury mechanisms (Koga et al., 2011; Koga et al., 2010). Moving up the kinetic chain, it is unclear how these kinematic and kinetic differences observed during landing and cutting relate to overall hip mechanics.

Sex differences in the mechanics of sports actions are commonly recognized (Kernozek et al., 2005; Schmitz et al., 2007; Weinhandl et al., 2016; Weinhandl et al., 2010). Females are reported to demonstrate kinematics, kinetics, muscle activations, and energy absorption strategies during landing and cutting that increase their risk for knee injury (Chappell et al., 2007; Malinzak et al., 2001; Pollard et al., 2004; Salci et al., 2004; Sigward and Powers, 2006; Zazulak et al., 2005). In a study exploring sex differences in unilateral and bilateral drop landings, females exhibited increased vertical ground reaction forces (GRF) and knee valgus compared to males during unilateral, but not bilateral conditions (Pappas et al., 2007a). Interestingly, during the same study no significant differences in hip kinematics were identified. Similar relationships in knee valgus behavior have been reported across various change of direction tasks (Jones et al., 2014; McLean et al., 2005) and single leg landing and cutting (Kristianslund and Krosshaug, 2013). In addition, females performed unilateral landings and relative height land-and-cuts with less hip flexion and knee adduction ROM (Weinhandl et al., 2015). In terms of neuromuscular control during landing, females are reported to demonstrate greater medial to lateral muscle activation imbalances (Palmieri-Smith et al., 2009) and land with greater quadricep activations (Shultz et al., 2009) compared to males. Given the biarticular structure of the hamstrings and rectus femoris, sex differences in neuromuscular control strategies may impact hip and knee biomechanics (Brown et al., 2014).

Examination of hip kinematics and loading during unilateral sport actions will assist in providing a foundation to begin exploring if proposed sex difference trends in knee mechanics persist within respect to the hip. Thus, the purpose of the current study was to explore sex differences in hip loading, GRF, and lower extremity kinematics during unilateral drop landings and land-and-cut tasks. It was hypothesized that land-and-cuts would incur larger peak forces and utilize less sagittal ROM compared to landing-only trials. Similarly, it was also hypothesized females would incur larger peak force and utilize less sagittal ROM compared to males.

Methods

Prior to data collection, ethics approval for experimental procedures was received from the university institutional review board and informed written consent was obtained from each participant. Forty-one participants (21 females; 20 males) from the university and surrounding communities volunteered for study participation. A general health history questionnaire was used to aid in screening participants with respect to inclusion and exclusion criteria. Inclusion criteria required that participants were: 18-30 years, recreationally active for at least 30 minutes 3 or more days per week, and participated in at least 1 activity per week including jumping and landing (i.e. basketball or volleyball). Exclusion criteria included: any history of lower extremity surgery, any injury to the lower extremities in the past six months, and presence of any condition that would impair the ability to jump and land safely.

Three-dimensional (3-D) marker coordinate data (200-Hz) and GRFs (2000-Hz) were captured during two landing tasks using a Vicon MX eight-camera motion analysis system

(Vicon, Centennial, CO, USA) and flush mounted Bertec force plates (Bertec, Columbus, OH, USA). The force plates and motion analysis system were calibrated according to manufacturer's specifications before data collections. For testing, all participants wore spandex shorts (and sports tops for females) and laboratory standard footwear (Air Max Glide, Nike, Beaverton, OR, USA). Eighteen individual retroreflective markers were placed at specified anatomical landmarks for calibration purposes (Weinhandl et al., 2010). Rigid plates with four retro-reflective markers in a cluster were placed on the torso, pelvis, and bilateral thighs, shanks, and feet for tracking in landing trials. A 3-s standing calibration trial was collected with the participant standing with one foot on each force plate and arms crossed high across the chest. Calibration-only markers were then removed.

Participants completed five trials of unilateral drop landings and land-and-cut maneuvers with their right leg. All landings were completed with a drop height of 40-cm. For the all landing maneuvers, participants began on top of the box while standing on their left leg. The right limb was lifted with hip and knees flexed approximately 90°. When ready, the participant stepped from the platform and either landed or immediately performed the cutting task. For landing-only tasks, the participant was required to land on the intended limb without additional instructions. For the land-and-cut task, the participant landed on the right limb and immediately cut to the contralateral side along a 1-m wide path oriented at 45° (Weinhandl et al., 2016). The 1-m wide path ensured participants maintained a cutting angle of 40°–50°. For both conditions, a successful trial required participants to perform the task without stepping down or jumping up from the box, land with the foot entirely on the force plate, and maintain crossed arms. Additionally, land-and-cut trials were required to perform the side cut without hesitation while staying within the boundaries. Landing tasks were counterbalanced across subjects. Testing sessions lasted approximately 30 minutes.

Data reduction and analysis was performed with Visual3D (v4.95, C-Motion Inc., Rockville, MD). Raw 3-D marker coordinate and GRF data were low-pass filtered using a 4th-order, Butterworth filter with a cutoff frequency of 10-Hz. Subject specific, eight-segment kinematic models (trunk, pelvis, and bilateral thighs, shanks, and feet) were created for each subject from calibration data. The model contained 19-dofs of freedom attributed to 3 rotations from the head-arms-trunk, 3 rotations and 3 translations from the pelvis, 3 rotations from the hips, 1 rotation from each knee, and 1 rotation from each ankle. Initial contact (IC) with the force plate was defined as the first timepoint vertical GRF exceeded 20N. Landing phases for both landing tasks were defined from IC to the first global minimum in knee extension (i.e. maximum knee flexion). The location of hip joint centers was determined using 25% of the distance from ipsilateral to contralateral trochanter markers (Bennett et al., 2016; Weinhandl et al., 2010). Knee joint centers were identified as half the distance between epicondyle markers (Grood and Suntay, 1983) and ankle joint centers were identified as half the distance between malleoli markers (Wu et al., 2002).

In OpenSim (Delp et al., 2007), the Gait2392 model was used with doubled strength and activation to simulate the landing phase. The inverse kinematics problem was solved at each frame to compute model joint angles using a least squares approach while accounting for constraint weights (Spoor and Veldpaus, 1980). Inverse dynamics were addressed using calculated joint angles and measured GRF data to identify net joint reaction forces and moments necessary to maintain equilibrium. Static optimization was used to estimate muscle forces at each time step using the sum of muscle stress squared as the optimization criteria (Anderson and

Pandy, 2001b). RMS reserve joint torques were verified to be below 5% of net joint moments (Hicks et al., 2015). The same inputs from static optimization were combined with generated muscle force values in the JointReaction analysis algorithm (Steele et al., 2012) to calculate hip joint forces (HJF). Hip joint torques from the JointReaction analysis outputs were randomly sampled trials were confirmed negligible.

Resultant GRF and HJF were calculated from 3-D components and used for all analyses. HJF were expressed in the femoral reference frame. Peak GRF (pGRF) and peak HJF (pHJF) were identified as the first global maximums after IC. Loading rates for pGRF and pHJF represent average loading from IC to the respective maximal measure. The impact phase was defined from IC to pGRF. Continuous resultant HJF, resultant GRF, as well as 3-D hip, knee flexion, and ankle dorsi/plantar-flexion measures from the landing phase were normalized to 101 data points. Averages for each continuous variable were calculated by sex and task.

Statistical analyses for discrete variables were performed via separate two by two (sex × task) analyses of variance (SPSS Inc., Chicago, IL, USA). Significance threshold was set *a priori* as $p < 0.05$. Kinetic discrete variables examined were pGRF, pHJF, pGRF loading rate, pHJF loading rate, and time to pGRF and pHJF. Kinematic discrete variables were 3-D hip, knee flexion, ankle planta/dorsi-flexion measures at IC, as well as, impact phase ROMs (ROM_I) and landing phase ROMs (ROM_L). Effect size was quantified using eta squared (η^2) and calculated as the sum of squares for each factor divided by the total sum of squares for sex, task, sex-task interaction, and error terms. Continuous variables were analyzed using custom MatLab (MathWorks, Natick, MA, USA) code for mean ensemble curve analysis with corresponding two standard deviation intervals. At each percentage of the normalized landing phase, areas of non-overlap were identified as significantly different.

Results

One participant was removed from analyses due to errors in data collection. The last two female participants to undergo data collection were also removed to maintain balanced sex groups. The final group was comprised of 19 females (height = 1.62 ± 0.05-m; mass = 60.04 ± 7.36-kg) and 19 males (height = 1.80 ± 0.05-m; mass = 79.13 ± 10.10-kg).

Continuous resultant forces and joint angles are provided in Figures 5-8. Mean ensemble curve analyses for all continuous variables did not identify areas of non-overlap between sexes. Significant differences in continuous forces were found between landing tasks for GRF between 20-28% of landing phase and for HJF between 21-25% of landing phase. Significant differences in joint angles were identified between landing tasks for hip adduction (42-100%), hip rotation (0-28%), and ankle dorsi/plantar-flexion (20-100%). There were no significant differences for continuous hip and knee flexion between landing tasks.

Discrete variable means ± standard deviations are provided in Table 2. There were no significant interaction effects identified for sex and landing task ($p<0.050$). Significant main effects for sex were identified for pGRF ($F_{1,72} = 4.776$, $p = 0.032$, $\eta^2 = 0.055$), time to pGRF ($F_{1,72} = 8.691$, $p = 0.004$, $\eta^2 = 0.125$), time to pHJF ($F_{1,72} = 5.178$, $p = 0.026$, $\eta^2 = 0.062$), IC knee flexion ($F_{1,72} = 12.018$, $p = 0.001$, $\eta^2 = 0.126$), ROM_I hip flexion ($F_{1,72} = 11.913$, $p = 0.001$, $\eta^2 = 0.134$), ROM_I hip adduction ($F_{1,72} = 23.533$, $p < 0.001$, $\eta^2 = 0.244$), ROM_I knee flexion ($F_{1,72} = 6.402$, $p = 0.014$, $\eta^2 = 0.082$), ROM_I ankle angle ($F_{1,72} = 6.506$, $p = 0.013$, $\eta^2 = 0.071$), ROM_L hip adduction ($F_{1,72} = 29.398$, $p <0.001$, $\eta^2 = 0.260$), and ROM_L hip rotation ($F_{1,72} = 4.276$, $p = 0.042$, $\eta^2 = 0.056$). Females landed with 0.20-BW greater pGRF without an accompanying significant increase in pHJF. Females used 6.45-ms more time to reach pGRF and 14.96-ms more time to reach pHJF. At IC, the only significant sex difference was 4.09° less knee flexion observed in females compared to males. For ROM_I, females utilized larger 3-D hip, knee flexion, and ankle dorsi/plantar-flexion ROMs compared to males. For ROM_L, females utilized 5.69° less hip adduction ROM and 2.19° more hip rotation.

Significant main effects for landing task we identified for pGRF ($F_{1,72}$ = 8.035 p = 0.006, η^2 = 0.092), pGRF loading rate ($F_{1,72}$ = 11.983, p = 0.001, η^2 = 0.141), time to pGRF ($F_{1,72}$ = 6.487, p = 0.013, η^2 = 0.125), IC hip rotation ($F_{1,72}$ = 11.640, p = 0.001, η^2 = 0.137), IC knee flexion ($F_{1,72}$ = 9.302, p = 0.003, η^2 = 0.097), ROM_I hip flexion ($F_{1,72}$ = 4.785, p = 0.032, η^2 = 0.054)), ROM_I ankle ($F_{1,72}$ = 12.332, p = 0.001, η^2 = 0.134), ROM_L hip adduction ($F_{1,72}$ = 11.111, p = 0.001, η^2 = 0.098), and ROM_L ankle ($F_{1,72}$ = 31.163, p < 0.001, η^2 = 0.289). Landing-only trials incurred 0.25-BW higher pGRF compared to land-and-cut trials. In addition, landing-only trials reached pGRF 5.57-ms faster and incurred a 5.57-BW/ms higher loading rate. At IC, landing-only trials were performed with 5.49° more hip abduction and 3.60° less knee flexion. Over the impact phase, landing-only trials also utilized 1.75° more hip flexion ROM_I and 8.59° less ankle dorsi/plantar-flexion ROM_I. Over the landing phase, landing-only trials utilized 3.49° more hip adduction ROM_L and 12.39° less ankle dorsi/plantar-flexion ROM_L compared to land-and-cuts.

Table 2. Groups means ± STD for study two discrete variables.

	Males	Females	Land	Land-and-Cut
Peak Force (BW)				
pGRF[#,*]	3.25 ± 0.37	3.45 ± 0.43	3.47 ± 0.43	3.22 ± 0.36
pHJF	10.06 ± 1.43	10.26 ± 1.99	10.40 ± 1.53	9.92 ± 1.90
Loading Rate (BW/s)				
pGRF	56.13 ± 13.05	57.92 ± 11.21	61.55 ± 11.16	52.50 ± 11.47
pHJF	165.57 ± 42.38	155.43 ± 63.75	170.32 ± 53.54	150.68 ± 53.37
Time to Peak Force (ms)				
pGRF[#,*]	57.63 ± 7.26	64.08 ± 11.85	58.07 ± 8.91	63.64 ± 10.92
pHJF[#]	59.34 ± 23.00	74.30 ± 33.46	63.99 ± 31.79	69.65 ± 27.15
IC (°)				
Hip Flexion	15.91 ± 8.48	16.64 ± 10.04	15.41 ± 8.45	17.14 ± 10.00
Hip Adduction	-15.98 ± 6.27	-16.80 ± 5.73	-15.98 ± 6.27	-16.33 ± 6.56
Hip Rotation[*]	-3.42 ± 7.93	-4.90 ± 7.00	-6.90 ± 6.80	-1.41 ± 7.16
Knee Flexion[#,*]	-18.62 ± 6.39	-14.53 ± 4.32	-14.78 ± 4.52	-18.38 ± 6.41
Ankle Dorsiflexion	-19.52 ± 10.92	-24.46 ± 12.33	-23.15 ± 12.34	-20.83 ± 11.35
ROM_I (°)				
Hip Flexion[#,*]	7.51 ± 2.69	10.28 ± 4.26	9.77 ± 4.14	8.02 ± 3.26
Hip Adduction[#]	4.12 ± 1.82	6.41 ± 2.26	5.50 ± 2.45	5.03 ± 2.24
Hip Rotation	6.26 ± 3.71	5.79 ± 2.94	6.00 ± 3.45	6.05 ± 3.26
Knee Flexion[#]	19.86 ± 3.48	22.86 ± 6.33	21.37 ± 4.83	21.35 ± 5.78
Ankle Dorsiflexion[#,*]	30.70 ± 9.14	36.94 ± 13.41	29.53 ± 9.36	38.12 ± 12.57
ROM_L (°)				
Hip Flexion	19.86 ± 7.56	21.22 ± 10.27	22.43 ± 8.59	18.66 ± 9.09
Hip Adduction[#,*]	10.23 ± 3.36	15.92 ± 6.00	14.82 ± 5.93	11.33 ± 4.73
Hip Rotation[#]	11.17 ± 5.72	8.98 ± 3.03	9.98 ± 5.62	10.17 ± 3.57
Knee Flexion	50.14 ± 7.50	51.14 ± 10.68	49.62 ± 8.37	51.66 ± 9.93
Ankle Dorsiflexion[*]	45.55 ± 10.26	49.98 ± 12.55	41.57 ± 10.02	53.96 ± 9.68

[#] Significant main effect for sex ($p < 0.05$)
[*] Significant main effect for landing task ($p < 0.05$)

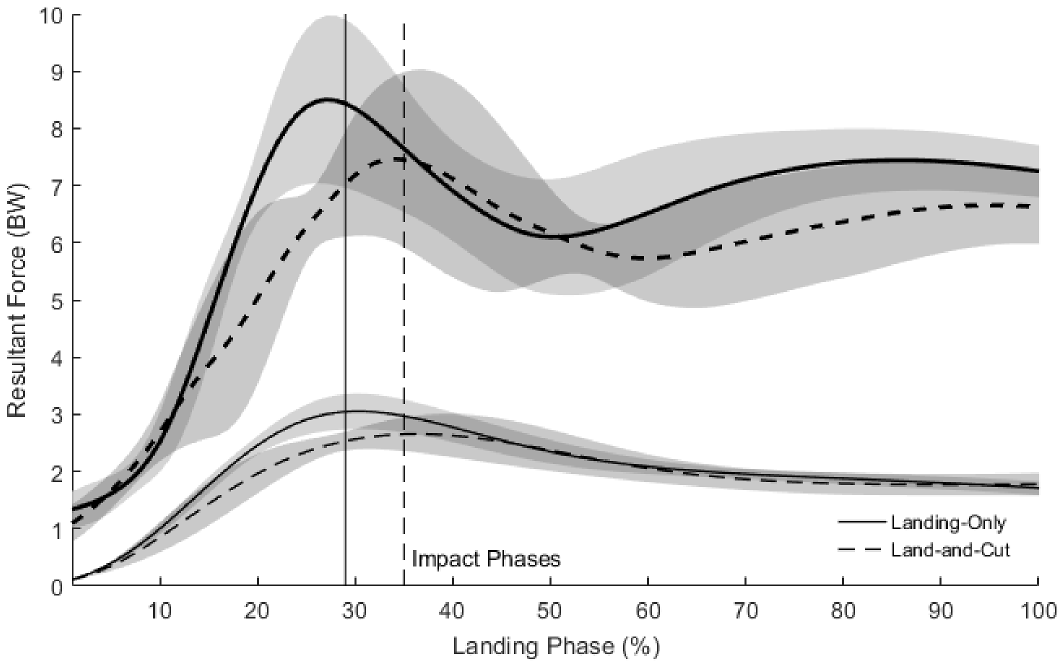

Figure 5. Study two continuous normalized group mean GRF and HJF (bold) ± two STD for landing task.

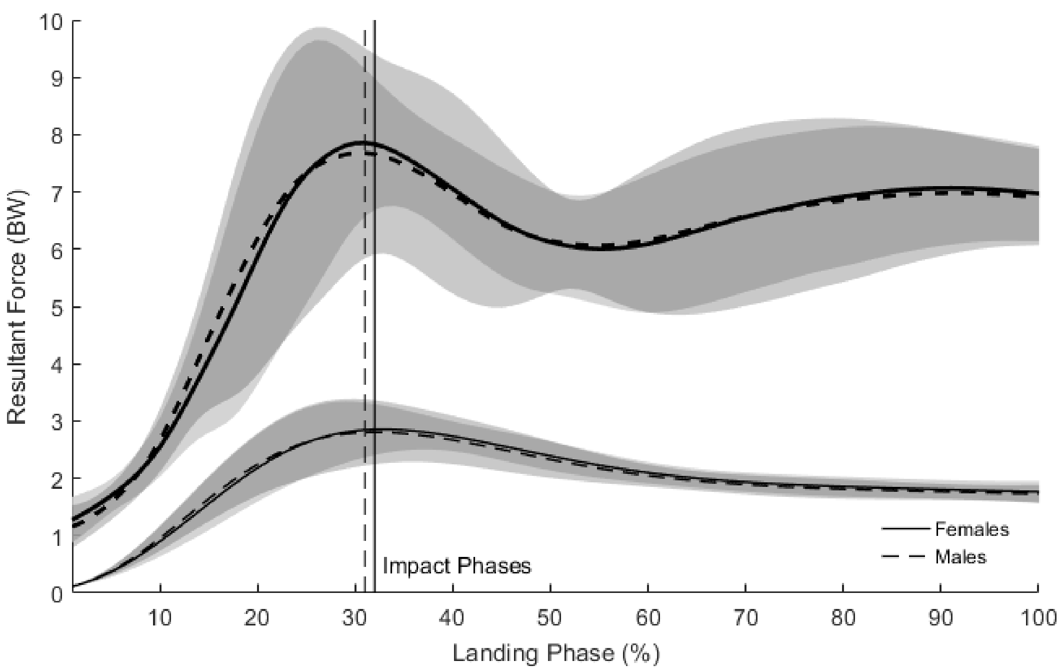

Figure 6. Study two continuous normalized group mean GRF and HJF (bold) ± two STD for sex.

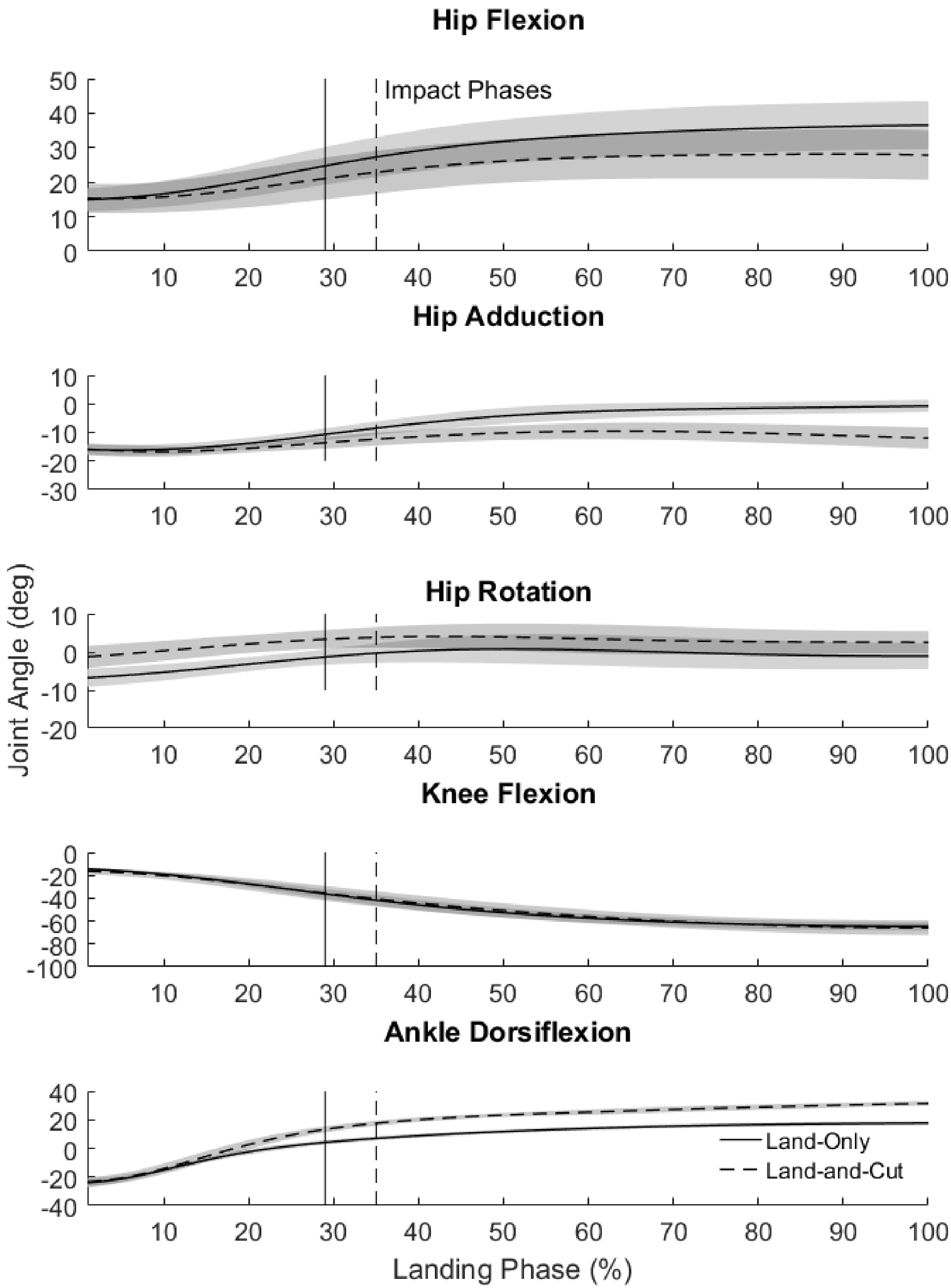

Figure 7. Study two continuous normalized group mean joint angles ± two STD for landing task.

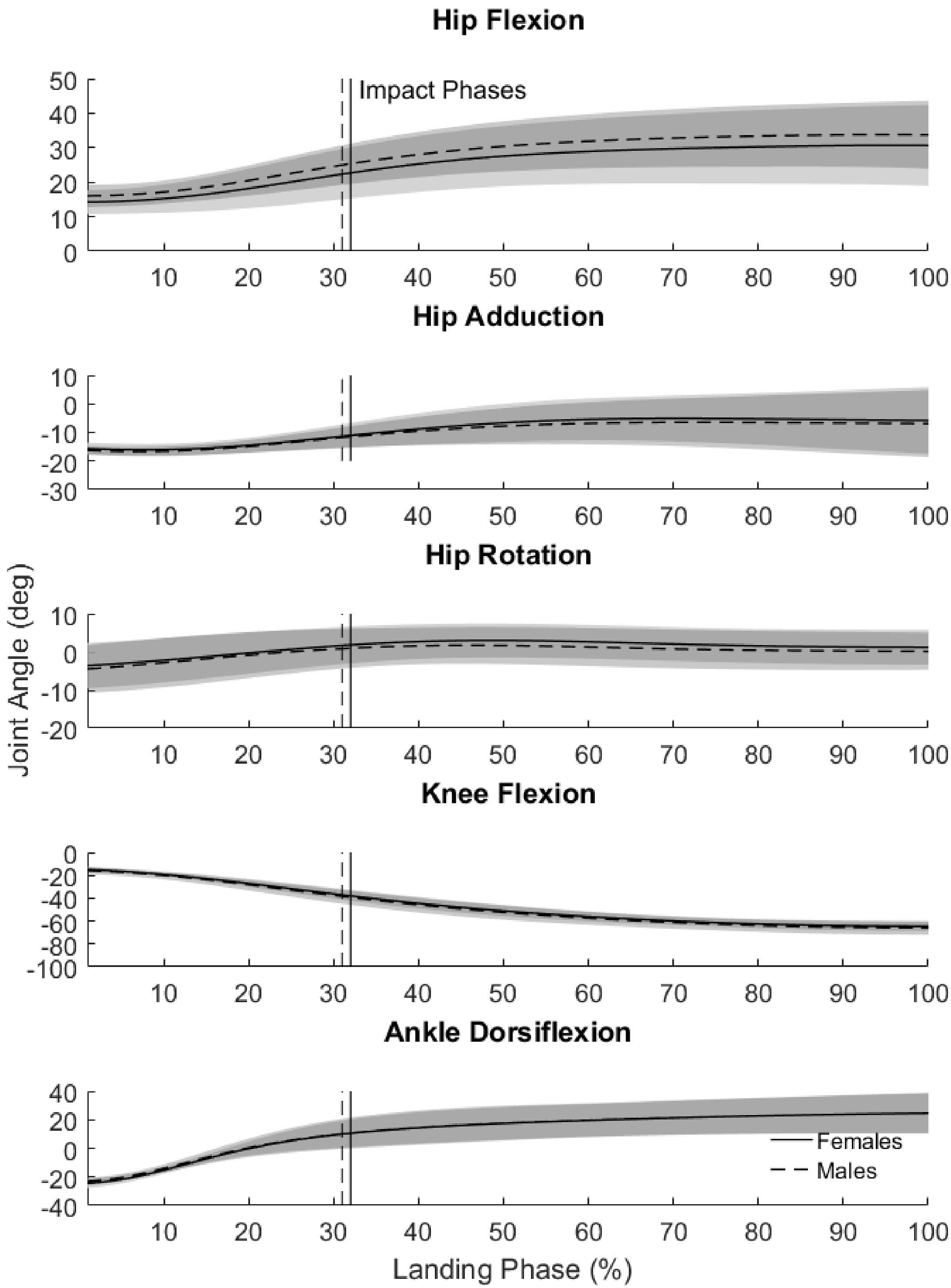

Figure 8. Study two continuous normalized group mean joint angles ± two STD for sex.

Discussion

It was hypothesized that land-and-cut tasks and females would generate greater pGRF and pHJF while exhibiting reduced sagittal ROM compared to landing-only tasks and males, respectively. The results partially supported these hypotheses. There was a significant increase in pGRF and pGRF loading rate with a reduction in time to peak pGRF for the landing-only task. However, these kinetic differences were not accompanied by differences in pHJF variables nor sagittal knee ROM_I or ROM_L. Other sagittal differences identified were greater ankle dorsi/plantar-flexion ROM_I and ROM_L and reduced hip flexion ROM_I during land-and-cut tasks. In the frontal plane, hip adduction ROM_L was greater in landing-only tasks. At IC, landing-only tasks were performed with more external hip rotation and less knee flexion compared to land-and-cut tasks.

In support of our hypothesis, females performed landings with significantly greater pGRF. However, the second part of the hypothesis was refuted as pHJF did not indicate a significant difference between sexes. Increased normalized peak vertical, posterior, and resultant GRFs have been identified in females during landing tasks (Kernozek et al., 2005; Pappas et al., 2007a; Weinhandl et al., 2015). Females took more time to achieve pGRF and pHJF than males. As a result, pGRF and pHJF loading rates differences were not identified. In examining the IC kinematics, males landed with more knee flexion compared to females. Weinhandl et al. (2015) showed similar, yet nonsignificant, sex difference trends in IC knee flexion during 40-cm unilateral landings. Pappas et al. (2007a) also identified similar IC group mean knee flexion angles (e.g. pooled sex data for landing task) using a unilateral landing from 40-cm, but they did not identify sex differences in IC knee flexion using a landing-only task. Females utilized a larger hip adduction ROM_L and less hip rotation ROM_L. Over the impact phase, females employed a larger ROM_I for hip flexion and adduction, knee flexion, and ankle dorsi/plantar-flexion. The increased hip adduction ROM observed in females has been well documented in sporting actions and considered a contributing factor in knee valgus collapse (Ford et al., 2006; Hewett et al., 2005; McLean et al., 2004; Myer et al., 2005).

While significant kinetic and kinematic differences were identified between sexes and tasks for discrete measures, only task differences were identified for continuous measures. Continuous GRF force tracings for both landing tasks were unimodal, which is consistent with previous unilateral landing task studies (Ali et al., 2014; Hargrave et al., 2003). Continuous HJF force tracings presented smaller second peaks which were more pronounced for the landing-only task. Landing-only continuous GRF and HJF were significantly higher during 20-28% and 21-25% of the landing phase, respectively. Interestingly, these ranges occur during the impact phase.

In the sagittal plane, continuous hip and knee flexion were not significantly different between tasks; however, land-and-cut tasks were performed with less ankle plantar-flexion from 20-100% of landing phase. Only ankle dorsi/plantar-flexion differences aligned with the differences in continuous GRF. From 20-28% of landing, landing-only tasks were performed with more internal hip rotation. Land-and-cut tasks experienced more hip abduction from 42-100% of landing. These differences in hip rotation and abduction seem logical given the task demands (i.e. the land-and-cut task was performed on the right leg with a cut to the left while the landing-only task was a simple drop landing on the right leg). However, the relative timing of hip rotation differences during the impact phase and hip adduction differences after the impact phase is interesting.

A lack of sex differences in the continuous variables was surprising, especially for GRF, and may have been impacted by the use of time normalization to the landing phase. The female impact phase ended later in the landing phase compared to males, which caused maximum GRF between sexes to occur at different landing phase percentages. The landing phase was selected for normalization as it captures the downward phase of landing and maximum knee flexion is commonly used to signify the end of landing (Decker et al., 2003; Yeow et al., 2010a). Examination of continuous kinetics and kinematics normalized over the impact phase may provide additional information on sex differences due to the high GRF and HJF loading rates.

These findings can be explored in terms of hip health with our current knowledge of FAI mechanics and etiology. FAI is a condition brought about by a lack of fit between the head of the femur and the acetabulum (Ganz et al., 2003). Anatomically, there is insufficient clearance between the bony structures to allow normal movement without creating bone-on-bone contact. Hip impingement is most likely to occur during motions incurring hip flexion with internal rotation (de Silva et al., 2016; Stull et al., 2011), with adduction furthering increasing approximation of the femoral neck and acetabulum (Ganz et al., 2008). Cutting sports are proposed to incur a higher prevalence than other types of sports (Nawabi et al., 2014). In our continuous analysis, land-and-cut tasks exhibited fairly neutral to slight internal hip rotation during the impact phase compared to a more externally rotated hip for landing-only. However, over the impact phase landing-only GRF and HJF were similar or higher than land-and-cut forces. Also over this phase, landing-only tasks used greater hip flexion ROM_I, pGRF, and pGRF loading rate. After the impact phases, the land-and-cut tasks presented more hip abduction than landing-only. This timing corresponds to a second nonsignificant peak in HJF observed for landing-only, but not land-and-cut. Yet, landing-only trials also presented greater hip adduction

ROM_L. In general, land-and-cut hip positioning over impact and landing phases may have demonstrated subtle positioning characteristics more likely to yield hip impingement, however resultant GRF and HJF did not indicate greater loading. Further analysis using 3-D force components may tease apart directional loading differences.

The data and results of the current study should be interpreted with respect to several limitations. First, the 'asymptomatic' participants in the current study were young adults (18 to 30-yrs), recreationally active, and free from lower extremity injuries or pain. As no diagnostic imaging or tests were performed, specific individual hip anatomies are not known and undiagnosed pathologies may exist. Thus, results can be only be generalized based on this sample population. Second, the selected tasks provided anticipated and constrained versions of sporting activities conducted in a laboratory setting. As an exploratory study, the landing tasks were selected to maximize standardization across data collections. Unanticipated landings may result in different landing mechanics (Brown et al., 2009). Third, the musculoskeletal model used was designed to focus on hip mechanics with 3-dof of hip motion and only 1-dof each for the knees and ankles. It is difficult to compare measured kinematics from the current study with existing studies employing knee and ankle models with additional dofs. Last, the landing heights used were absolute and not related to individual participants jumping ability. The use of absolute heights may have represented different task demands between participants (Weinhandl et al., 2015).

Increasing awareness of hip specific pathologies, particularly FAI, has highlighted a paucity in our understanding of asymptomatic hip mechanics during dynamic tasks. Moving forward, a well-developed understanding of asymptomatic hip mechanics in young adults, during sport actions will provide a basis of comparison to identify potentially risky biomechanical trends. The purpose of this study was to begin exploring sex and hip task differences during landing-only and land-and-cut maneuvers. To this end, the study utilized continuous measures of kinematics and kinetics to capture the entire landing phase, as well as discrete measurements from IC and peak knee flexion. Females and landing-only tasks produced greater pGRF, decreased time to pGRF, greater ROM_L hip adduction, greater ROM_I hip flexion, and decreased IC knee flexion compared to males and land-and-cut tasks. However, ankle ROM_I was higher in females compared to males, while it was lower in landing-only tasks compared to land-and-cuts. In addition, females utilized larger ROM_I for hip flexion, hip adduction, knee flexion, and ankle plantar/dorsi-flexion, as well as, a smaller ROM_L for hip adduction. Despite the increased ROMs employed by females, significant differences in pHJF were reported. Kinematic and kinetic differences for sex and task indicate

CHAPTER 6: SEX AND LANDING HEIGHT EFFECTS ON HIP JOINT LOADING, GROUND REACTION FORCES, AND LOWER EXTERMITY KINEMATICS

Introduction

Femoroacetabular impingement (FAI) represents a significant and growing cause of disability in otherwise asymptomatic active adolescents and adults. Diagnoses of symptomatic and asymptomatic FAI are rapidly increasing (Colvin et al., 2012; Montgomery et al., 2013; Sampson, 2005). The incidence of impingement pathomorphologies are estimated as high as 10% to 15% of the general population and believed responsible for 22% to 55% of all hip pain (Groh and Herrera, 2009; Schilders et al., 2009). The number of hip arthroscopies performed between 2004 and 2009 increased approximately 365%, with the majority of patients between 20-yrs and 39-yrs of age (Montgomery et al., 2013). While short term results and return to play statistics are optimistic, the long-term impact on development of hip osteoarthritis (OA) remains to be seen. Our understanding of hip biomechanics during activities of daily life and athletics in steadily improving; however, there is much we do not fully grasp. Traditionally, hip conditions have been associated primarily with older, less active adults. With a strong link emerging between athletic participation and impingement morphology, we need a more comprehensive understanding of asymptomatic, hip mechanics in action.

Several authors have proposed increased risks of modified landing mechanics and injury potential are associated with increasing landing height (McNitt-Gray, 1993; Weinhandl et al., 2015; Yeow et al., 2010a; Yeow et al., 2009). According to unilateral landing study findings, body weight normalized GRFs increases with landing height (Ali et al., 2014; Yeow et al., 2010a) and the largest kinematic differences are related to frontal plane motion at the hip and knee (Kernozek et al., 2005). Kinematic differences have been reported in terms of hip and knee positioning at initial contact, maximum angles, and ROMs. However, when GRFs were assessed with additional normalization for height, no significant differences were reported between vertical and posterior GRF and landing height (Weinhandl et al., 2015).

A reduction in knee and ankle energy absorption has been observed with landing height increases, while hip energy absorption remained similar. The shift in lower extremity energy absorption with increased landing height did not include a shift toward proximal joints (Montgomery et al., 2014; Weinhandl et al., 2015). Proximal-to-distal energy absorption strategies are believed to expose passive structures to higher forces when landing (Norcross et al., 2010). Diminished strength in the female hip musculature in conjunction with altered joint positioning have been proposed as risk factors for ACL injuries (Decker et al., 2003; Khayambashi et al., 2016; Lawrence et al., 2008). However, more energetically hip dominant landing mechanics proposed in males (Yeow et al., 2009) may result in altered hip joint loading and long term implications not yet fully realized. Examining hip kinematics and kinetics during dynamic, sporting activities would garner a more complete understanding of lower-extremity mechanics and provide a basis of comparison for pathological presentations. While direct measurement of hip joint forces is extremely invasive, musculoskeletal modeling can be used to estimate joint forces using measured kinematics and kinetics.

Unilateral landings are a common movement in sporting activity and associated with an increased injury prevalence compared to bilateral landings (Ali et al., 2014). In addition, unilateral landings present different mechanics than those reported in bilateral ladings (Ali et al., 2014; Dufek and Bates, 1990). As such, the purpose of the current study is to examine what, if any, sex differences exist in hip joint loading, ground reaction forces (GRF), and lower extremity kinematics at different unilateral drop landing heights.

Methods

Eighty-three asymptomatic, physically active participants between the ages of 18 and 35 were recruited from a medium sized urban university and the surrounding communities. Upon arrival, all participants were given a full explanation of the study purpose and procedures. Written informed consent was obtained in accordance with university institutional review board polices. A general health history questionnaire was provided to screen for study inclusion criteria. Inclusion criteria stated participants would be recreationally active, defined as participation in at least 30 minutes of exercise per week including a minimum of one jumping related activity per week, have no history of lower extremity surgery, no lower extremity injury within the past six months, and no condition that would impair their ability to jump or land.

Three-dimensional (3-D) marker coordinate data were collected for landing trials at 200-Hz using an eight-camera Vicon MX motion analysis system (Vicon, Centennial, CO, USA) with ground reaction forces (GRF) measured synchronously at 2000-Hz using two flush mounted force plates (Bertec, Columbus, OH, USA). Prior to data collection, the motion capture system and force plates were calibrated according to manufacturer's specifications. For testing, participants wore snug fitting spandex shorts and lab standard tennis shoes (Air Max Glide, Nike, Beaverton, OR, USA). Single retro-reflective markers were placed on the skin over specific anatomical landmarks with special adhesive tape for calibration purposes only (Weinhandl et al., 2010). Calibration-only markers were placed bilaterally on the acromioclavicular joints, iliac crests, greater trochanters, medial and lateral knee epicondyles, medial and lateral malleoli, first metatarsal heads, and fifth metatarsal heads. Marker tracking in movement trials was employed using four-marker, rigid cluster plates positioned on the upper torso and pelvis, as well as, bilateral thighs, shanks, and feet (Weinhandl et al., 2016). A three-second static trial was collected with the participant standing motionless, arms crossed high over the chest, and each foot on a separate force plate. Calibration-only markers were removed before landing trials.

Participants completed five successful unilateral landings on their right limbs from 30-cm and 50-cm. Selected heights are absolute and commonly used in drop landing research (Bruton et al., 2013). For the unilateral drop landing, participants begin standing on the contralateral limb desired for landing with intended landing limb in approximately 90° of hip and knee flexion. When ready, the participant stepped away from the platform and landed on the force plate with the intended landing limb. For a successful landing, the participant performed the task without stepping down or jumping up from the box, landed with their entire foot on the force plate, landed without hopping or jumping, landed without the contralateral limb contacting the ground, and maintained arms crossed high across the chest. Practice trials were allowed. Instructions were not provided on landing technique. Landing height order was counterbalanced between participants.

An eight-segment kinematic model was created in Visual 3D (v4.95, C-Motion Inc., Rockville, MD) based upon standing calibration data using a least squares approach (Spoor and Veldpaus, 1980). Raw 3-D marker coordinate and GRF data were low-pass filtered using a 4th-order, Butterworth filter with a cutoff frequency of 10-Hz (Kristianslund et al., 2012). Trochanter hip joint centers were defined as 25% of the distance from ipsilateral to contralateral greater trochanter markers (Bennett et al., 2016; Weinhandl et al., 2010). Joint centers for the knees and ankles were defined as the midpoints between medial and lateral epicondyles (Grood and Suntay, 1983) and malleoli markers (Wu et al., 2002), respectively. The start of landing was defined as initial contact (IC) with the force plate, specifically the time point where GRF exceeded 20N. The landing phase was defined from IC to the maximum knee flexion (MF).

Subject specific models were incorporated for use with Gait2392 in OpenSim (SimTK, 2016) to simulate each landing phase (Delp et al., 1990b). Gait2392 is an 8-segment model with 19 degrees of freedom (dof) actuated by 92 Hill-type contractile elements in series with tendon (Delp et al., 1990a). The segments represented the head-arms-trunk (3 rotations), pelvis (3 translations and 3 rotations), hips (3 rotations each), knees (1 rotation each) and ankles (1 rotation each). Each dof was accompanied by a reserve actuator that was manually checked after simulations were complete. Muscle forces were estimated using static optimization at each time step with inverse kinematics outputs, external load data, a scaled model and the sum of muscle stress squared as the optimization criteria (Anderson and Pandy, 2001b; Crowninshield and Brand, 1981a; Kaufman et al., 1991a). Hip joint forces (HJF) were calculated using the JointReaction analysis algorithm within OpenSim. To assist in model validation, resulting hip torques were verified as null, reserve actuators were checked to be below 5% of RMS net joint moments, and values were compared to published modeling data (Hicks et al., 2015). The HJFs represent the internal loads associated with all unmodeled structures and are expressed with respect to the pelvis on the femur, in the femoral reference plane, and acting through the joint center (Steele et al., 2012).

Resultant GRF and HJF were calculated from 3-D force components and normalized by body weight. All GRF and HJF discussed are in reference to resultant measures except when stated otherwise. Variables were classified as discrete or continuous. Continuous variables were taken from the group mean for each percentage of the normalized landing phase (0-100%). Continuous sex and condition group means for GRF, HJF, 3-D hip ROM, knee flexion ROM, and ankle dorsi/plantar-flexion ROM were calculated. Discrete kinematic variables were 3-D hip, knee flexion, and ankle dorsi/plantar-flexion ROM during the impact (ROM_I) and landing phases (ROM_L), as well as, joint measures at IC. The impact phase was defined from IC to the first peak in GRF (pGRF). Discrete kinetic variables were pGRF, pGRF timing, peak HJF (pHJF), pHJF timing, GRF loading rate, and HJF loading rate. pGRF and pHJF were defined as the first peaks in respective resultant plots. Loading rates for pHJF and pGRF were calculated by the respective force measure divided by time to peak value. All discrete variables were checked for normality and homogeneity prior to statistical analyses. IC represents the starting time for all analyses.

Two by two analyses of variance (sex x height) were performed for each discrete variable (SPSS Inc., Chicago, IL, USA) ($p<0.050$). Effect size was quantified using eta squared (η^2) and calculated as the sum of squares for each factor divided by the total sum of squares for sex, height, sex-height interaction, and error terms. Continuous variables were compared using mean ensemble curve analyses, with corresponding two standard deviation intervals at each percentage of the normalized landing phase. Statistically different areas of the landing phase were identified where the two standard deviation intervals of the mean ensemble curves between sexes and heights did not overlap.

Results

Three participants' data were not included in analysis. One participant was excluded due to errors in data collection. The last two female participants to undergo data collection were excluded to maintain balanced sex groups. Forty females (height = 1.64 ± 0.06-m, mass = 61.19 ± 8.66-kg) and 40 males (height = 1.79 ± 0.08-m, mass = 79.78 ± 13.53-kg) were used for data analysis. Five trials were collected for each condition; however, only the first three successful trials were used for analysis.

Continuous GRF and HJF are provided in Figures 9 and 10. There were no significant differences between sex and any of the continuous kinetic or kinematic variables. Significant differences in continuous forces between landing height were identified for GRF between 1-29% and HJF between 14-24% and 77-100% of the landing phase. The only joint angle with significant differences associated with landing height was hip adduction from 1-22% of the landing phase (Figure 11).

Means ± standard deviation for all discrete variables are provided in Table 3. There were no interaction effects between landing height and sex (p<0.050). Significant main effects for sex were identified for pGRF ($F_{1,156}$ = 8.553, p = 0.004, η^2 = 0.036), pHJF ($F_{1,156}$ = 8.894, p = 0.003, η^2 = 0.044), GRF loading rate ($F_{1,156}$ = 4.301, p = 0.040, η^2 = 0.019), IC hip flexion ($F_{1,156}$ = 7.312, p = 0.008, η^2 = 0.045), ROM_I hip flexion ($F_{1,156}$ = 7.607, p = 0.007, η^2 = 0.045), ROM_I hip adduction ($F_{1,156}$ = 23.919, p ¸0.001, η^2 = 0.095), ROM_I knee ($F_{1,156}$ = 10.189, p = 0.002, η^2 = 0.046), ROM_L hip rotation ($F_{1,156}$ = 7.988, p = 0.005, η^2 = 0.046), ROM_L hip adduction ($F_{1,156}$ = 44.434, p < 0.001, η^2 = 0.133). Females landed with 0.19-BW higher pGRF, 0.92-BW higher pHJF, and 3.62-BW/s higher pGRF loading rate compared to males. As a ratio, these values represent 5.5% and 9.9% increases respectively in pGRF and pHJF for females. Only hip flexion differed at IC, with females exhibiting 5° more hip flexion than males. For ROM_I, females exhibited 1.54° greater hip flexion ROM, 1.75° greater hip adduction ROM, and 1.56° greater knee flexion ROM. Females exhibited 5.46° more hip adduction ROM_L and 2.01° less hip rotation ROM_L.

Significant main effects for landing height were identified for pGRF ($F_{1,156}$ = 76.517, $p <$ 0.001, $\eta^2 = 0.316$), pHJF ($F_{1,156}$ = 35.502, $p < 0.001$, $\eta^2 = 0.177$), pGRF loading rate ($F_{1,156}$ = 62.764, $p < 0.001$, $\eta^2 = 0.281$), pHJF loading rate ($F_{1,156}$ = 8.454, $p = 0.004$, $\eta^2 = 0.051$), time to pGRF ($F_{1,156}$ = 13.754, $p < 0.001$, $\eta^2 = 0.111$), IC hip adduction ($F_{1,156}$ = 77.410, $p < 0.001$, $\eta^2 =$ 0.331), ROM_I hip flexion ($F_{1,156}$ = 5.066, $p = 0.026$, $\eta^2 = 0.030$), ROM_I hip adduction ($F_{1,156}$ = 70.229, $p < 0.001$, $\eta^2 = 0.280$), ROM_I hip rotation ($F_{1,156}$ = 7.625, $p = 0.006$, $\eta^2 = 0.046$), ROM_I knee ($F_{1,156}$ = 10.189, $p = 0.002$, $\eta^2 = 0.059$), ROM_L hip flexion ($F_{1,156}$ = 14.287, $p < 0.001$, $\eta^2 =$ 0.083), ROM_L hip adduction ($F_{1,156}$ = 129.852, $p < 0.001$, $\eta^2 = 0.390$), ROM_L hip rotation ($F_{1,156}$ = 9.864, $p = 0.002$, $\eta^2 = 0.056$) , ROM_L knee ($F_{1,156}$ = 19.489, $p < 0.001$, $\eta^2 = 0.111$), and ROM_L ankle ($F_{1,156}$ = 4.372, $p = 0.027$, $\eta^2 = 0.027$). Landings from 50-cm incurred 0.57-BW more pGRF and 1.86 BW more pHJF compared to 30-cm landings, representing 17.4% and 21.1% increases respectively. Loading rates from 50-cm measured 13.82-BW/s higher for pGRF and 28.46-BW/s higher for pHJF. Only hip adduction differed at IC, with 50-cm landings exhibiting 8.44° more hip abduction than 30-cm landings. For all angles except ankle dorsi/plantar-flexion, ROM_I were greater for 50-cm than 30-cm landings. All ROM_L were greater for 50-cm landings compared to 30-cm landings.

Table 3.
Group means ± STD for study three discrete variables.

	Males	**Females**	**30-cm**	**50-cm**
Peak Force (BW)				
pGRF$^{\#,*}$	3.47 ± 0.49	3.66 ± 0.52	3.28 ± 0.43	3.85 ± 0.43
pHJF$^{\#,*}$	9.26 ± 2.34	10.18 ± 1.98	8.79 ± 1.91	10.65 ± 2.11
Loading Rate (BW/s)				
pGRF$^{\#,*}$	58.97 ± 12.76	62.59 ± 13.23	53.87 ± 10.45	67.69 ± 11.76
pHJF*	126.92 ± 65.59	140.74 ± 60.65	119.60 ± 59.42	148.06 ± 64.32
Time to Peak Force (ms)				
pGRF*	60.21 ± 7.93	59.83 ± 6.86	62.15 ± 7.94	57.96 ± 6.17
pHJF	73.19 ± 33.91	79.10 ± 47.65	77.44 ± 41.20	74.85 ± 41.69
IC (°)				
Hip Flexion$^{\#}$	8.61 ± 10.41	13.32 ± 11.50	11.70 ± 11.06	10.22 ± 11.33
Hip Adduction*	-18.42 ± 7.68	-18.36 ± 7.09	-14.17 ± 5.74	-22.61 ± 6.33
Hip Rotation	-5.68 ± 8.52	-4.53 ± 6.05	-5.83 ± 6.14	-4.38 ± 8.43
Knee Flexion	-15.81 ± 6.19	-15.40 ± 5.72	-16.46 ± 6.23	-14.76 ± 5.55
Ankle Dorsiflexion	-24.68 ± 11.38	-26.99 ± 7.50	-25.26 ± 10.03	-26.41 ± 9.33
ROM_I (°)				
Hip Flexion$^{\#,*}$	6.94 ± 3.18	8.48 ± 3.89	7.08 ± 3.35	8.34 ± 3.80
Hip Adduction$^{\#,*}$	4.70 ± 2.44	6.45 ± 2.96	4.08 ± 2.00	7.08 ± 2.78
Hip Rotation*	6.41 ± 3.35	5.72 ± 2.84	5.40 ± 2.88	6.73 ± 3.21
Knee Flexion	20.73 ± 4.32	22.29 ± 3.88	20.50 ± 4.46	22.52 ± 3.62
Ankle Dorsiflexion$^{\#,*}$	32.39 ± 9.26	32.93 ± 8.49	31.91 ± 9.05	33.41 ± 8.66
ROM_L (°)				
Hip Flexion*	17.31 ± 8.76	19.23 ± 10.24	15.53 ± 8.12	21.01 ± 10.11
Hip Adduction$^{\#,*}$	13.04 ± 5.65	18.86 ± 8.15	11.46 ± 5.00	20.80 ± 6.65
Hip Rotation$^{\#,*}$	10.87 ± 5.47	8.86 ± 3.54	8.75 ± 3.79	10.97 ± 5.26
Knee Flexion*	48.64 ± 11.27	48.41 ± 9.22	45.12 ± 9.75	51.93 ± 9.67
Ankle Dorsiflexion*	44.32 ± 1.60	45.20 ± 7.94	43.23 ± 9.18	46.30 ± 9.32

$^{\#}$ Significant main effect for sex ($p < 0.05$)
* Significant main effect for landing height ($p < 0.05$)

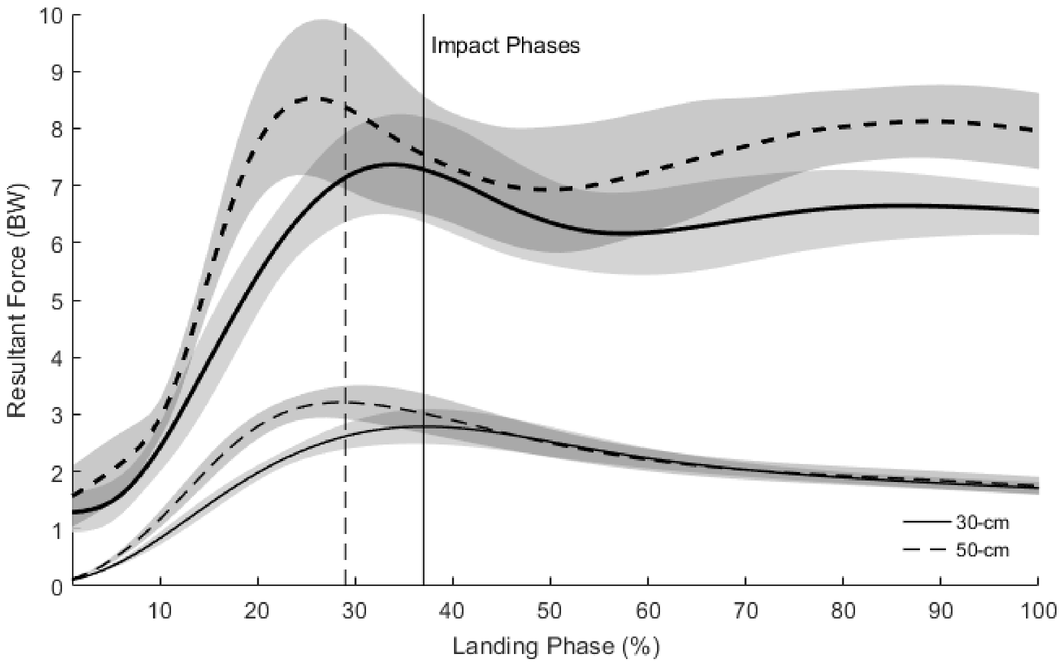

Figure 9. Study three continuous normalized group mean GRF and HJF (bold) ± two STD for landing height.

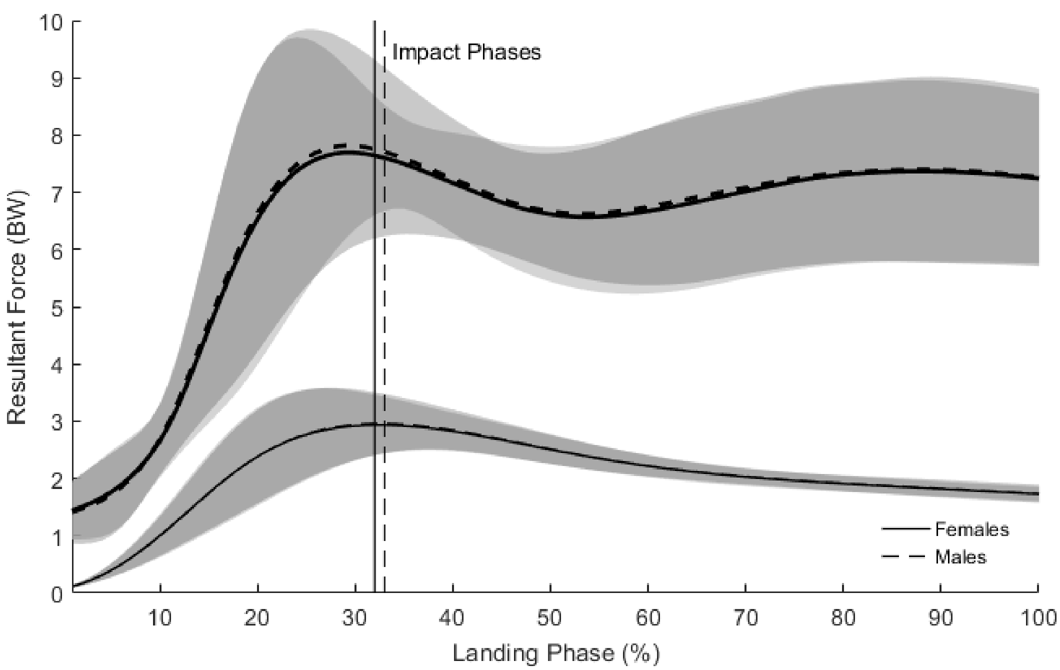

Figure 10. Study three continuous normalized group mean GRF and HJF (bold) ± two STD for sex.

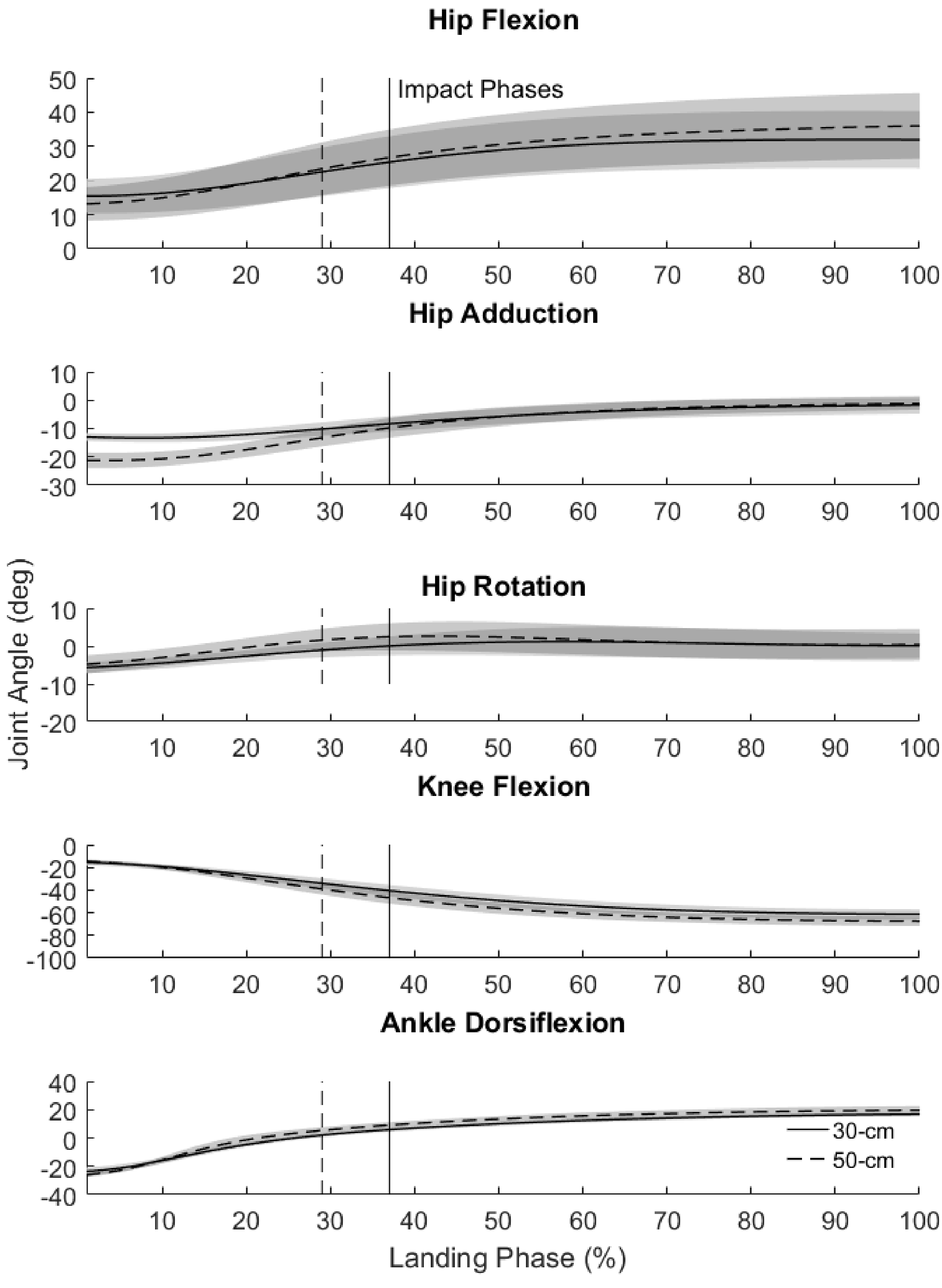

Figure 11. Study three continuous normalized group mean joint angles ± two STD for landing height.

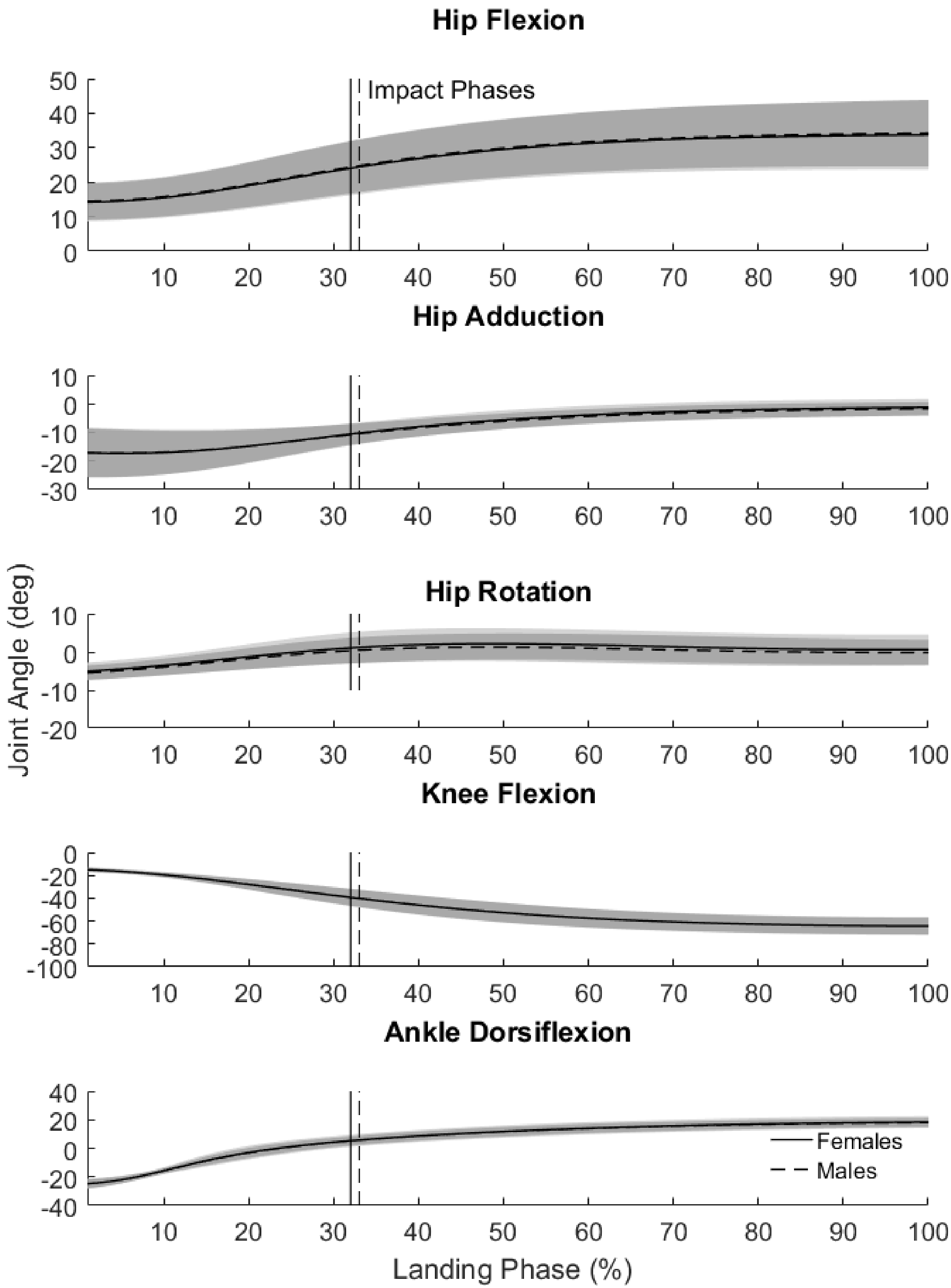

Figure 12. Study three continuous normalized group mean joint angles ± two STD for sex

Discussion

The purpose of the current study was to examine the effects of sex and landing height on unilateral HJF, GRF, and lower extremity kinematics. Kinematic and kinetic variables were assessed using data from discrete time points, as well as continuous data over the landing phase. Due to the exploratory design of this study, these variables were selected to quantify differences over the duration of the landing phase in an effort to tease apart sex and landing height differences. The HJF calculated in the current study are higher than direct measurement via instrumented prosthesis (Bergmann et al., 2001; Bergmann et al., 1993). However, differences in sample characteristics and the task employed may have been a factor. Currently, HJF have not been modeled during a dynamic landing task. The values calculated here are comparable to previous modeling studies examining walking and running (Giarmatzis et al., 2015; Rooney and Derrick, 2013).

Examination of the continuous kinetic and kinematic variables did not illuminate any significant sex differences. However, landing height differences were identified for continuous GRF, HJF, and hip adduction. GRF were higher for the 50-cm landings for 0-28% of the landing phase which corresponds to the impact phase for 50-cm landings. The GRF curves for both heights were unimodal which is consistent with unilateral landings (Ali et al., 2014; Dufek and Bates, 1990; Hargrave et al., 2003). However, the 50-cm height reached the end of the impact phase faster, with force curve shifted to the left (e.g. earlier in the landing phase). This shift played a large role in the early differences in GRF between heights. Meanwhile, the HJF curve demonstrated height differences from 13-23% and 77-100% of the landing phase with the 50-cm landings yielding higher forces. The initial range occurs during the impact phase and shows similar unimodal characteristics as the GRF tracing. After the impact phase, HJF behavior differs between heights with the 50-cm landing incurring higher forces again as the landing phase ends. This behavior was not observed in the GRF tracings. The only continuous kinematic difference identified with respect to landings was associated hip adduction early in the impact phase. Landings from 50-cm demonstrated more hip abduction beginning at 0% or IC, which persisted through 21% of the landing phase.

Increasing landing height from 30-cm to 50-cm resulted in alterations of all kinetic measures and ROMs with the exception of time to pHJF and ankle plantar/dorsi-flexion ROM_I. Conversely, landing height differences in IC kinematics did not demonstrate the same behavior. Only hip adduction at IC demonstrated differences between heights. Previous research employing similar landing heights reported differences in IC hip flexion and knee adduction, with hip adduction demonstrating a sex-landing height interaction (Weinhandl et al., 2015). The current study did not identify any sex-landing height interactions. A positive relationship between unilateral landing height and knee flexion was also identified by Ali et al. (2014) and Fagenbaum and Darling (2003). Yeow et al. (2010a), on the other hand, identified similar behavior for double leg landings, but not for bilateral landings. Ali et al. (2014) did not identify significant main landing height effects for peak ankle plantar flexion or peak hip flexion. However, the study used different definitions of the landing phase and reported peak values where ROMs were employed here.

The increased pGRF reported is consistent with previous research on unilateral and bilateral landing heights (Ali et al., 2014; Seegmiller and McCaw, 2003; Yeow et al., 2009; Zhang et al., 2000). pHJF demonstrated a similar increase with landing height. The hip joint must assist in dissipating GRF incurred while landing, thus increases in HJF with reported higher GRFs are not surprising. Interestingly, time to pHJF was not significantly different between heights, while time to pGRF was different. 3-D hip and knee flexion ROM_I were larger for the 50-cm landings compared to 30-cm; however, ROMs were not calculated over the hip joint loading phase and cannot be quantitatively compared.

Although sex differences were not detected in continuous variables, analysis of discrete measures identified kinematic and kinetic differences. Females incurred higher pGRF and pHJF compared to males. Higher posterior (Weinhandl et al., 2015) and vertical (Pappas et al., 2007a) GRFs in unilateral landings have been previously reported in females. The increase in female pHJF was in excess of four times the change observed in pGRF. Females also incurred a higher pGRF loading rate, without a significant increase in pHJF loading rate. The times to pGRF and pHJF did not demonstrate a sex effect. Kinematic differences between sexes were identified in the all three planes of hip motion and knee flexion. Females presented with more hip flexion at IC and went on to utilize larger hip and knee flexion $ROM_{I.}$ Sagittal plane joint motion has been associated with GRF loading. In the current study, females exhibited larger pGRF and pGRF loading rate in conjunction with larger sagittal ROM_I. Increased ankle ROM in females has been proposed to help absorb energy at the ankle and safely dissipate GRFs (Decker et al., 2003; Kernozek et al., 2005). It should be noted, over the landing phase there were no differences in sagittal plane hip, knee, or ankle $ROM_{L.}$ Females utilized a larger hip adduction ROM over both impact and landing phases. Only an increase in hip rotation ROM_I was associated with males.

The observed sex differences suggest there are kinematic and kinetic differences associated with hip joint loading during a unilateral landing. Modified hip joint loading has been proposed as an etiological factor in FAI. Females incurred higher pHJF, however over the entire landing phase HJF were similar between sexes. The highest HJF were detected during the impact phase during which time females utilized more hip flexion and adduction ROM. The increased female hip ROM may help limit excessive loading in the same joint configuration and decrease the development of pathomorphologies. Sports associated with large amounts of internal hip rotation have been proposed to incur an increased risk of developing FAI, as such, it is important to note that hip rotation ROM did not demonstrate sex differences. Limitations in hip ROM are associated with symptomatic FAI and typically attributed to pathomechanical constraints from boney contact and soft tissue edema (Ganz et al., 2003). Future research may examine if limited hip ROM is a prexisting factor in the etiogenesis of impingement morphology. Individuals that perform landings with a more limited hip flexion and adduction ROM from a young age may create loading conditions more susceptible to physiological maladaptations.

There are several limitations to consider when interpreting the study findings. First, the drop heights used are absolute and may present different task demands depending on individual jumping ability (Weinhandl et al., 2015). Vertical jump height was not assessed and it is possible the 50-cm height represented an unlikely landing height for some participants. Second, only right leg landings were included in analysis. Previous findings have failed to identify clear bilateral differences with respect to left/right or dominant/nondominant limbs (Van der Harst et al., 2007b). Third, the anatomical model used for analysis was defined and scaled using retroreflective spherical markers adhered to the skin. These markers are used to approximate locations of boney anatomical landmarks, many of which have complex structures. Potential sources of error during data collection are related to proper and consistent placement of markers, as well as, marker movement artifact during data collection. Errors incurred during the placement and movement trial data collection phases are propagated throughout analysis (Hicks et al., 2015). These placement errors were minimized by the use of an experienced clinician for all marker placement. Movement artifact errors were reduced via marker clusters for tracking movement trials.

Less than ideal distribution of GRFs have been proposed as mechanisms of ankle and knee injuries (Boden et al., 2009); however, the implications for hip health are unknown. Hip mechanics of sporting actions have received limited attention due in part to the low incidence of acute hip injuries. High forces across the hip are a proposed etiological factor in the development of FAI and OA, yet what 'high forces' means with respect to specific mechanical characteristics is vague. It is difficult to classify pathologically high forces without first generating a range associated with asymptomatic, normal hips. Musculoskeletal modeling of hip joint forces allows for the non-invasive estimation of joint loading during dynamic tasks. The model employed was individually scaled for each participant, but did not require medical imaging. The use of such model allows for generalized comparisons between large numbers of participants with a minimum of computational cost. The current study modeled 80 asymptomatic, active participants and identified sex and landing height differences. In summary, increases in landing height yielded similar increases in pGRF and pHJF with increased 3-D hip and knee flexion ROMs during impact and landing phases. Females demonstrated increased pGRF and pHJF, increased hip adduction ROM_I and ROM_L, increased hip and knee flexion ROM_I, and decreased hip rotation ROM_L.

CHAPTER 7: SUMMARY AND CONCLUSIONS

Increased forces between the femur and acetabulum are a proposed mechanism for damage in traditional hip OA and FAI (Ganz et al., 2003). Males have been reported to land with a hip dominant landing strategy that is associated with a reduced risk of knee injury (Weinhandl et al., 2015). Conversely, females are associated with increased GRF during landings and sporting actions (Pappas et al., 2007a; Van der Harst et al., 2007a). The resulting hip loading associated with these landing strategies are not fully understood. Unfortunately, hip joint forces cannot be directly measured via non-invasive methodologies. These forces can be estimated through musculoskeletal modeling using measured kinematics and kinetics (Bergmann et al., 2001; Delp et al., 2007). The overall purpose of this dissertation was to explore the effects of sex, limb side, landing task, and landing height on landing hip mechanics. The problem was addressed via three specific studies intended to examine differences in hip joint forces, ground reaction forces, and lower extremity kinematics between (1) sexes and limb sides, (2) sexes and landing tasks, and (3) sexes and landing height. The first study failed to identify significant bilateral differences. As such, bilateral comparisons were not performed for the second or third studies. Sex differences, on the other hand, were identified in the first study and persisted across the second and third studies.

The data from 80 participants (40 females, height = 1.640-m, mass = 61.190 kg; 40 males, height = 1.793-m, mass = 79.777-kg) were included for analysis in this dissertation. All participants were between the ages of 18 and 30-yrs, asymptomatic, recreationally active, had no history of lower extremity surgery, nor any condition that would impair their ability to land from a jump. Participants were fitted with retroreflective markers for calibration and tracking purposes. Once complete, a series of landing trials were performed with conditions counterbalanced across subjects. Trial conditions consisted of unilateral drop landings on each limb from 30-cm, 40-cm, and 50-cm box heights. In addition, a 38-participant subset (19 females, height = 1.625-m, mass = 60.041-kg; 19 males, height = 1.796-m, mass = 79.125-kg) of this sample performed land-and-cut tasks using the 40-cm box height. Measured kinematics and kinetics were to used simulate the landing phase via individually scaled, 3-D musculoskeletal models. Each musculoskeletal model was used to calculate internal joint loads experienced at the hip. This data was used in conjunction with measured ground reaction forces and lower extremity kinematics to describe landing mechanics. Landing mechanics were classified by landing and impact phases. Landing phase described the time from initial contact to maximum knee flexion, while impact phase describes the time from initial contact to maximum normalized resultant ground reaction force.

Several similarities in sex differences were identified across the three studies. In all three studies, females incurred larger normalized peak GRF. However, only the landing height study identified a corresponding increase in GRF loading rate or peak HJF in females. Females performed landings with more hip adduction and less hip rotation ROMs during the landing phase in all three studies. Also across studies, females utilized more ROM during the impact phase for hip flexion, hip adduction, and knee flexion. When landing task was included in assessment, females performed the impact phase with more ankle plantar/dorsi-flexion ROM as well. These findings highlight potential differences in how the hip undergoes loading during a unilateral jump landing with respect to resultant forces and joint motion. Force application across differing ROM would potentially result in modified physiological responses. The larger hip flexion and adduction ROMs identified in females across conditions likely results in a different loading profile across the joint. Given the prevalence of FAI in male athletes may be higher than females and less active individuals, these differences warrant further investigation.

Landing condition differences were examined between landing limb side (study one), landing task (study two), and landing height (study three). In study 1, bilateral differences were not identified between right and left limbs during a 40-cm unilateral drop landing. These results suggest participants performed right and left sided unilateral drop landings with similar mechanics. When considering these findings with respect to current FAI research, 20% and 75% of individuals present with radiographic evidence of impingement morphology bilaterally (Allen et al., 2009; Ganz et al., 2008; Klingenstein et al., 2013). The lack of limb side differences may indicate both sides undergo similar joint loading that results in similar physiological changes.

In study 2, unilateral drop landings were compared with unilateral land-and-cuts. Only right leg landings were used. The landing-only task yielded increases in peak GRF and peak GRF loading rate with a reduction in time to peak GRF. No differences were detected with respect to peak HJF, peak HJF loading rate, or time to peak HJF. At initial contact, landing-only trials displayed increased external hip rotation and decreased knee flexion compared to land-and-cut trials. During the impact and landing phases, landing-only trials went on to exhibit reduced ankle plantar flexion ROM. Additional kinematic differences were identified with landing-only trials demonstrating increased hip flexion ROM over the impact phase and increased hip adduction ROM over the landing phase. Participation in cutting sports have been posited as a factor in development of FAI in young athletes due to repetitive hip loading with flexion and internal rotation (Clohisy et al., 2013; de Silva et al., 2016; Siebenrock et al., 2011). The landing-only condition was performed with a more externally rotated hip and increased hip flexion ROM during the impact phase compared to the land-and-cut. It is possible that the combination of landing-only kinematics without a task difference in HJF brings about loading conditions less likely to elicit physiological changes associated with FAI.

In study 3, unilateral drop landings were compared from 30-cm and 50-cm heights. Again, only right limb data was used for analysis. Landing from increased height resulted in greater peak values, loading rates, and time to peak GRF and HJF. At initial contact, 50-cm landings exhibited more hip abduction than those from 30-cm. Increased ROM was observed for 3-D hip and knee flexion during impact and landing phases. Conversely, 50-cm ankle plantar/dorsi-flexion ROM was greater during the landing phase, but not the impact phase. The increased forces reported with landing from the higher height were not surprising given the additional kinetic energy needed to be dispersed.

Interestingly, no interaction effects between sex and landing condition were identified for all kinematic and kinetic variables examined. One possible interpretation of these findings is that while males and females may have responded differently to a given landing condition, they experienced similar modifications in response to changing landing conditions. Another important factor to consider when interpreting reported findings is the sensitivity of Gait2392 in modeling of landing tasks. Joint forces were estimated based on calculations of muscle excitations-activations-forces which generated the best fit for the specified cost function. As such, there exists a range of possible solutions to the mathematical expression that would have generated different muscle activation patterns. Given muscle actions are directly related to joint loading, true differences in mechanics may have been masked due to discrepancies between estimated and actual muscled activations.

These findings present a vast array of future research paths to continue unraveling the complex relationships associated with hip mechanics. This study only grazed the surface with respect to the kinematic and kinetic relationships of unilateral landings. Joint moments and energetics would provide more information about how the lower extremity musculature dissipates ground reaction forces. Inclusion of electromyography would further aide in capturing the timing of muscle activation and addition of isometric testing would further individualize the musculoskeletal model. Moving forward, studies which focus on specific populations may illuminate subtle differences with sporting activies given the increased incidence of hip injury in cutting sports. Prospective studies of adolescent athletes landing mechanics have been in place for the past decade to track ACL injuries. Similar studies could be designed to identify trends in landing mechanics which may eventually led to hip pathologies.

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VITAE

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