SEX DIFFERENCES IN LOWER LIMB BIOMECHANICS DURING A SINGLE-LEG CUT WITH BODY BORNE LOAD

by

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A thesis

submitted in partial fulfillment

of the requirements for the degree of

Master of Science in Kinesiology, Biophysical Studies

Boise State University

May 2018

BOISE STATE UNIVERSITY GRADUATE COLLEGE

DEFENSE COMMITTEE AND FINAL READING APPROVALS

of the thesis submitted by

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Thesis Title: Sex Differences in Lower Limb Biomechanics during a Single-Leg Cut with Body Borne Load.

Date of Final Oral Examination: 23 March 2018

The following individuals read and discussed the thesis submitted by student AuraLea Carylon Fain, and they evaluated her presentation and response to questions during the final oral examination. They found that the student passed the final oral examination.

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ABSTRACT

Introduction: Musculoskeletal injuries are ever-increasing in military personnel, particularly females. These musculoskeletal injuries are attributed to adaptations in lower limb biomechanics while performing routine military tasks, such as a single-leg cut, with the addition of body borne load. However, it is unknown if females and males exhibit similar lower limb biomechanics with the addition of body borne load during these tasks. This study sought to compare the lower limb biomechanical adaptations exhibited by females and males performing a single-leg cut with body borne load. **Methods:** Eleven females and 17 males had lower limb biomechanics quantified during a single-leg cut with four body borne load conditions (*20, 25, 30 and 35 kg*). Each participant performed five successful cuts off each limb (*dominant and non-dominant*). **Statistical Analysis:** For analysis, initial contact (IC) and peak stance (PS) hip, knee and ankle 3D rotations and PS moments, and peak proximal tibial shear were calculated. Each variable was submitted to a RM ANOVA to test main and interaction effects of sex (*male, female*), load (*20, 25, 30 and 35 kg*), and limb dominance (*dominant vs. non-dominant*). **Results:** Body borne load increased peak proximal anterior tibial shear force ($p = 0.011$). However, females exhibited significantly greater proximal tibial shear with the 25 kg configuration compared to the 20 kg configuration ($p = 0.028$), while males exhibited greater peak proximal tibial shear force with 35 kg configuration compared to 20 kg ($p = 0.04$) and 25 ($p = 0.011$) kg configurations. During the cut, females exhibited significantly greater IC and PS hip adduction angle ($p = 0.016$ and $p = 0.015$), and PS hip adduction ($p < 0.001$) and knee external rotation ($p = 0.004$) moments compared to males. Males exhibited significantly greater PS hip flexion moment ($p = 0.041$) and knee flexion moment – but only with the 25 kg ($p = 0.04$) and 30 kg ($p = 0.022$) load configurations – compared to females. **Conclusion:** The addition of body borne load increases risk of musculoskeletal injury for military personnel performing a single-leg cut. Females exhibited hip and knee biomechanics reported to increase dynamic valgus loading of the knee and may have a greater risk of musculoskeletal injury during the single-leg cut compared to their male counterparts.

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ACL	Anterior cruciate ligament
Dom	Dominant
F	Female
Non	Non-dominant
M	Male

CHAPTER ONE: INTRODUCTION

Musculoskeletal injuries are an extensive and ever-increasing problem for the military. In 2012, military personnel suffered 2.2 million musculoskeletal injuries with associated healthcare costs exceeding 700 million dollars[1]. These musculoskeletal injuries commonly occur during basic and/or advanced training[2], and are the most prominent cause of long-term disability and medical discharge for military peronnel[3]. A majority of these musculoskeletal injuries, i.e. bone and soft tissue damage and disorders, occur in the lower limb[4–6] wherein the knee is the most commonly injured joint[7,8]. These injuries are attributed to maladaptive lower limb biomechanics that military personnel adopt during training[5,9]. Military activities require that personnel don body borne loads during all training and mission related tasks[10,11]. These body borne loads are typically composed of personal protective and fighting equipment and supplies[4] that routinely exceed 45 kg during military operations[6,12,13]. The addition of body borne load reportedly produces the maladaptive lower limb biomechanical patterns that increase the risk of sustaining a lower limb musculoskeletal injury during military activities [5,11]. However, the specific maladaptive lower limb biomechanical adaptations that occur with small incremental increases of body borne loads commonly worn during military training and mission related tasks (e.g. 20 kg to 35 kg) is largely unknown.

Body borne load reportedly results in decrements of physical performance[5] and increases risk of musculoskeletal injury for military personnel[6,11]. This reduction in performance and increase of injury risk is attributed to specific lower limb biomechanical adaptations adopted during the performance of locomotor tasks common to military training, such as running and cutting [10,11,14]. During these locomotor tasks, the addition of body borne load results in a significant increase in the peak vertical ground reaction force the musculoskeletal system must attenuate[15–17]. To safely attenuate these ground reaction forces and successfully execute each maneuver, military personnel increase stance time, and alter lower limb joint angles and joint moments[18,19]. For instance, military personnel exhibited a significant reduction in peak hip and knee flexion angle during military-relevant running and cutting tasks with a 40 kg compared to 20 kg body borne load[14,18,20]. This extended lower limb is thought to prevent collapse of the limb, but aids the transfer of the elevated ground reaction forces to the musculoskeletal system of the limb[21], subsequently increasing the risk of bony[22,23] and/or soft-tissue injury[24]. Further increasing the risk of musculoskeletal injury during military related tasks are the elevated joint moments that are adopted with the addition of body borne load. During a single-leg cut with body borne load, Brown et al.[14] reported that military personnel exhibited a significant increase in peak hip and knee flexion, and hip adduction joint moments. These elevated joint moments require greater force generation by the hip and knee musculature to prevent collapse of the limb, but increase the stresses placed on the musculoskeletal system and risk of injury[25]. Both large knee flexion, with an extended knee[26], and abduction moments reportedly strain the soft-tissue structures of the joint[27], leading to damage and injury. Yet, to date, it is unknown if military personnel exhibit maladaptive adaptations in lower limb biomechanics, and increased risk of musculoskeletal injury, when performing a single-leg cut with the small incremental changes of body borne load (i.e. 5kg, 10 and 15 kg) commonly worn during military related tasks.

During military training, female personnel are two and a half times more likely than their male counterparts to suffer a musculoskeletal injury[3,28]. This discrepancy in injury rate may be attributed to the sex dimorphism in lower limb biomechanics exhibited during locomotor activities routinely performed during military training, such as single-leg cutting. During unloaded cutting, females reduce peak hip and knee flexion joint angle[29,30], and exhibit greater hip adduction, and knee flexion and abduction joint moments compared to their male counterparts[31,32]. The extended limb posture adopted by females prevents collapse of the lower limb and allows for successful performance of the task, but may lead to their increased risk of musculoskeletal injury. Female's extended lower limb posture results in greater proximal tibial shear and subsequent loading of the joint's soft-tissues[23,33,34]. Despite their increased risk of musculoskeletal injury, it is currently unknown if females exhibit and/or exaggerate a sex dimorphism in lower limb biomechanics when performing a single-leg cut with body borne load. Previously, neither Krupenevich et al.[35] or Silder et al.[36] reported a significant difference in lower limb biomechanics between male and female participants when walking with body borne load, despite the fact females reportedly exhibit a sex dimorphism in hip and knee biomechanics during unloaded walking[37–39]. However, it is unknown if female exhibit a sex dimorphism in lower limb biomechanics during performance of rapid change of direction tasks routinely required of military personnel during training[40]. Considering females were recently given the opportunity to fill infantry positions in the U.S. Military[41], where they are routinely required to run and cut with body borne load, it is imperative to quantify and compare the lower limb biomechanical adaptions they exhibit with male personnel during performance of these locomotor tasks with the standardized body borne loads common to military activities. This work seeks to fill that critical void and examines the effect of small changes of body borne load (5 kg, 10 kg and 15 kg) on male and female participants during a single-leg cut.

Specific Aims:

Specific Aim 1:

To examine lower limb biomechanical adaptations exhibited during a single-leg cut with body borne loads (20, 25, 30 and 35 kg) commonly worn during military activities.

Hypothesis 1

Each small incremental increase of body borne load (*5kg, 10 kg and 15 kg*) will result in significant adaptation of lower limb biomechanics.

Subhypothesis 1.1

During the single-leg cut, participants will significantly decrease initial contact and peak of stance hip and knee flexion, hip adduction and knee abduction joint angles with the 5 kg, 10 kg and 15 kg increase in body borne load.

Subhypothesis 1.2

During the single-leg cut, participants will exhibit a significant increase in peak hip and knee flexion, hip adduction and knee abduction joint moments with the 5 kg, 10 kg and 15 kg increase in body borne load.

Significance

Determining the specific lower limb biomechanical adaptations that occur during a single-leg cut with small incremental increases of body borne load will help the military reduce the likelihood of training-related musculoskeletal injury associated with load carriage. Understanding the maladaptive lower limb biomechanics that occur as a result of common military body borne loads can guide injury prevention protocols and inform the design of military load carriage requirements and equipment.

Specific Aim 2:

To compare the lower limb biomechanical adaptations between male and female participants performing a single-leg cut with body borne loads (20, 25, 30 and 35 kg) commonly worn during military activities.

Hypothesis 2

With each small incremental increase of body borne load (*5kg, 10 kg and 15 kg*), female participants will exhibit larger adaptations in lower limb biomechanics during the single-leg cut compared to their male counterparts.

Subhypothesis 2.1

With a 10 kg and 15 kg increase in body borne load, females will exhibit a significant difference in initial contact and peak stance hip and knee flexion, hip adduction and knee abduction joint angles compared to the male participants. However, no significant differences in hip and knee joint angles will be evident between the male and female participants with the 5 kg increase in body borne load.

Subhypothesis 2.2

During the single-leg cut, females will exhibit larger peak hip and knee flexion, hip adduction and knee abduction joint moments, and proximal tibial shear with the 5 kg, 10 kg and 15 kg increase in body borne load compared to male participants.

Subhypothesis 2.3

During the single-leg cut, females will exhibit larger peak vertical ground reaction forces and more medially directed frontal plane GRF vector with the 5 kg, 10 kg and 15 kg increase in body borne load compared to male participants.

Significance

Determining if male and female participants exhibit differences in lower limb biomechanics during the performance of a single-leg cut task with body borne load can help the military tailor training and injury prevention protocols to the sex of the military personnel, and subsequently reduce the risk of musculoskeletal injury, particularly for female personnel.

CHAPTER TWO: LITERATURE REVIEW

This section aims to detail load carriage, specifically the 1) significance of military load carriage, 2) lower limb biomechanics of load carriage, 3) sex dimorphism in lower limb biomechanics, and 4) sex dimorphism in load carriage biomechanics.

Military Load Carriage

Load carriage, specifically locomotion while transporting an external, added mass supported on the upper torso or body by shoulder straps and/or hip belt[12], is commonly required of military personnel. It is often required that military personnel complete dynamic locomotor tasks, such as running or cutting, with the addition of these torso-borne loads ranging from 25 to 35 kilograms during training[41].

In the last half century, the body borne load military personnel are required to carry has increased 15.5 kg[3,6,11]. This additional load leads to a decrease in mobility[42], and likely the ever-increasing rate of musculoskeletal injury and medical disability among military personnel[3,43]. In fact, the rate of musculoskeletal injury, such as injury or damage to the bones and soft tissues of the lower limb (i.e. stress fracture or tendonitis), for military personnel has increased sevenfold in the last twenty-five years[3] with upwards of 60% of military personnel suffering a musculoskeletal injury during basic and advanced training[44]. These musculoskeletal injuries have both a significant financial and physical cost. Treating musculoskeletal injuries costs the armed services $5.5 billion dollars annually[43] and still routinely leads to medical discharge and long-term disability.

Aerobic fitness level has been linked to risk of musculoskeletal injury obtained during training[3], but is only one factor in the risk of musculoskeletal injury in military personnel. The increased incidence of musculoskeletal injury for military personnel [3,4,45–47] is most commonly attributed to lower limb biomechanics exhibited during completion of military-related tasks with the addition of load, as the bones and soft tissues of the lower limb are most commonly effected[5,7,46,47] and nearly half of these injuries occur at the knee[7]. Many of these injuries that occur during training can be defined as overuse injuries. Overuse injuries are the result of repetitive, weight-bearing and cyclical activity and most commonly occur in the lower limb and are the result of the inability of the musculoskeletal system to adapt to the stresses being put on the body[48,49]. Common overuse injuries are stress reactions, stress fractures, ligament and tendon injuries and joint injuries and the risk of their acquisition is exacerbated by the addition of a body borne load [49,50]. Furthermore, one third of these lower limb musculoskeletal injuries are reported to during mission-related training[51], where military personal routinely run, cut and march with load.

Load Carriage Biomechanics

Running

Running with body borne load results in significant spatiotemporal and lower limb kinematic and kinetic adaptations. How humans move while running is defined as the gait cycle and has separate phases, one of which is stance phase. Stance phase requires that the support limb acts to accept the weight of the body and associated forces and generate power to produce forward propulsion of the body from initial contact to toe off of the support limb foot. The addition of body borne load while running results in a significant increase of time of the gait cycle dedicated to the stance phase of gait [18,19,52–54]. This increase of stance time is a strategy of the load carrier to commit more time to the safe attenuation of the concurrent significant elevation in ground reaction force experienced by the body that is a consequence of an increase of body borne load[15–17,19,55]. The high ground reaction forces that occur due to the increase of body born load result in higher values of shock on the musculoskeletal system and are linked to boney injury acquisition[56,57]. Since there is an increase of ground reaction force, it stands to reason that there would be a subsequent increase in the magnitude of external frontal and sagittal plane loading as these ground reaction force resultant vectors are calculated using components of 3D ground reaction forces[58]. In an unloaded running condition, military personnel with a history of injury experience significantly higher magnitude external loading than their historically healthy counterparts. The increase in the magnitude of these vectors may increase risk of musculoskeletal injury in military personnel, but, to date, no studies have examined the influence of load on frontal and sagittal plane external loading vectors. In an effort to minimize these ground reaction forces acting on the body, military personnel donning body borne loads decrease stride length[59], or the distance between heel strikes of the same foot, while running. This spatiotemporal adaptation to load helps reduce the detrimental high ground reaction forces acting on the body, injury risk[54,60], and metabolic cost associated with load carriage[4]. However, even with a reduction of stride length, Willy et al. reported that the addition of body borne load leads to an increase of patellofemoral and tibiofemoral contact forces[34,54] and tibial stress[34] while running, both of which are linked to musculoskeletal injury[61,62].

The elevation of ground reaction forces acting on the body, and subsequent reduction in stride length, is paired with a significant reduction in joint motion while running with load[19]. While hip and knee flexion angles are higher at initial contact and peak stance with the addition of load, there is a significant reduction in the amount of change in the joint angle, or range of motion, which is a characteristic of a stiff limb[18,19,52]. Similarly, the Xu et al. reported that, while body borne load failed to elicit kinematic adaptations at the knee, the hip joint experienced significantly less range of motion[34], increasing limb stiffness while running with a load weighing 30% of participant body weight. The presence of increased limb stiffness may be an attempt of the body to minimize physiological cost associated with vertical displacement of the body's center of mass[4,63]. However, it aids in the transfer of the dangerous elevated ground reaction forces, increasing strain on the musculoskeletal system and injury risk.

Brown et al. did not find similar adaptations to hip and knee flexion angles at initial contact or peak stance during steady state running[64]. While load reportedly failed to produce any kinematic adaptations at the hip or knee, the addition of 20 and 40 kg body borne loads resulted in an increase of external flexion moments at both joints while running[18]. These extended limb postures when paired with high magnitude moments may prevent limb collapse, but further increase strain on the musculoskeletal system by aiding the transfer of high ground reaction forces up the kinetic chain of the body[25]. Further, high magnitude moments are indicative of an increase in mechanical work of the musculoskeletal system that occurs in an effort to prevent limb collapse. Work completed by the musculoskeletal system is defined power produced over time, with positive work characterizing power production and negative work characterizing energy absorption.

The addition of load results in an increase of total work completed across the limb[52], caused by an increase in both positive and negative work performed at the knee with the addition of load body borne load[65]. While these kinetic patterns are indicative of the increased reliance on knee joint to absorb and produce energy during stance[53], similar adaptations to load are not noted at the hip. The addition of load results in a significant reduction in positive work and an increase of negative work at the hip, evidence of decreased power generation in preparation for propulsion, but increased reliance on the hip for energy absorption. These increases of power production and energy absorption at the hip and the knee increase mechanical strain on structures of the musculoskeletal system, therefore elevating risk of injury and metabolic cost[4].

Single-leg Cutting

Cutting is defined a sudden change of direction, requiring rapid deceleration from the forward motion in order to move laterally. The nature of a single-leg cut task requires an increase of frontal and transverse plane motions and moments[66,67] in the support limb compared to running in an effort to successfully complete the maneuver and generate enough propulsion to power the body laterally. Due to the increase of load carriage requirements in military personnel and the associated increase in injury, it is imperative to understand if the addition of a body borne load exaggerates cutting mechanics that are linked to musculoskeletal injury as a non-contact mechanism (i.e. single-leg cut task) is the leading cause of soft tissue injury at the knee[68]. However, there is currently a dearth of research that has examined the influence of body borne load on lower limb biomechanics during a dynamic cutting maneuver.

As in running, Brown et al. reported that the addition of load results in a significant increase of time dedicated to stance during a single-leg cut task[14]. Further, it was reported that the addition of 6, 20 and 40 kg loads while cutting resulted in significant reductions in hip and knee flexion angles[14]. Additionally, increasing body borne load during a single-leg cut task results in a significant reduction of hip adduction[14]. These kinematic adaptations are paired with an increase of hip and knee flexion and hip adduction moments. As previously mentioned, the increase in joint moment magnitude is an indicator of mechanical work being done by the muscles to prevent limb collapse due to the increase of external load on the body. The presence of high magnitude moments when paired with angular reductions, as documented with the addition of load by Brown et al., not only increase strain on the musculoskeletal system, but aid in transfer of high ground reaction forces. As in loaded running, there is an additional attempt to attenuate the high ground reaction forces acting on the body, the energy absorption strategy adapts to load: the hip increases its contribution to energy absorption while the knee increases energy generation while performing a single-leg cut task[14]. The hip increasing its contribution to energy absorption is an indication of the body requiring more energy absorption than the knee musculature alone can supply.

Sex Dimorphism in Lower Limb Biomechanics

Running

Sex differences of lower limb biomechanics are well documented in both kinematic and kinetic variables of running gait cycle. When running, females are documented to have more range of motion in the frontal and transverse planes at both the hip and the knee[23,37,39,69–72]. Specifically, females exhibit significantly higher values of hip adduction and internal rotation, knee abduction and external rotation[23,73]. These kinematic patterns are indicative of an increase of valgus loading at the knee joint[74] and characterize female gait as unstable[75], increasing risk of soft-tissue injury at the knee in females during unloaded running. Greater hip joint flexion values are documented in female runners throughout stance phase[23,39] when compared to males, but this difference may be attributed to elevated values of anterior pelvic tilt documented in female runners[39,69]. Conversely, more extended knee joint postures are documented in females when compared to their male counterparts[70,76].

Elevated hip flexion values in female runners may also be attributed to sex-specific energy generation and absorption patterns. Females exhibit higher power peaks at the hip, indicative of greater energy absorption and generation at the hip, compared to males who rely more heavily on the knee and ankle joints to power their gait cycle[39]. Furthermore, female runners exhibit greater magnitudes in hip adduction[39] and flexion[23] moments in when compared to their male counterparts. Though these high magnitude moments paired with a more extended limb knee of the female runner may prevent limb collapse, as previously stated, they aid in the transfer of ground reaction forces, further increasing risk of musculoskeletal injury incidence in female runners[23,24,77].

Ground reaction force free moment is the torsional moment between the foot and the ground, and high values of this measure are thought to be linked to musculoskeletal injury, specifically stress fractures[78,79]. When examining ground reaction force free moment in female runners, it has been documented that females with history of stress fracture experience significantly higher magnitude of ground reaction force free moment compared to a healthy population[80]. However, it is unknown how the magnitude of these moments compare to that of male runners.

Single-leg Cutting

Female athletes are up to 10 times more likely to incur an musculoskeletal injury at the knee than their male counterparts[81]. This increase of injury risk may be linked to the documented sex dimorphism that females exhibit while completing a single-leg cut task. When cutting, it has been observed that females exhibit lower hip and knee flexion angles when compared to their male counterparts[29,82,83]. Furthermore, there is an reduction in hip peak abduction angles[29,84]. This reduction in hip abduction is paired with elevated knee valgus[29,85] and hip internal rotation[86], both of which lend the soft tissue structures of the knee to higher risk of rupture. Differences in Q-angle between males and females has been used as an explanation as to why females may have an increased risk of injury. The Q-angle is the angle measured from the anterior superior iliac spine to the middle of the patella. This angle is, on average, larger in females as they generally have a wider pelvis[87,88]. Beaulieu et al. measured the Q-angle of participants in an effort to see if anatomical variances may be, in part, responsible for sex dimorphism during a single-leg cut task, but found no significant differences between the sexes[89]. For this reason, it is unlikely the skeletal geometry plays a role in biomechanical differences documented in a single-leg cut task.

Extended joints in females performing a single-leg cut are paired with higher hip adduction[31] and internal rotation, and knee abduction[90] moments and lower internal peak hip extensor and knee flexor moment magnitudes[82,91]. These mechanical adaptations to a sudden change of direction are evidence of elevated knee valgus loading and increased transfer of ground reaction forces across the joints in females completing a single-leg cut, elevating their risk of injury acquisition when compared to mechanics exhibited by males.

Women in the Military

In December 2015, the Direct Ground Combat Assignment Rule was repealed[92], allowing female personnel to fulfill frontline, infantry positions previously unavailable to them. While the availability of these physically demanding positions is a new development, it is well-documented that there is a discrepancy of injury rates between male and female military personnel, with females at a much higher risk of incurring a musculoskeletal injury than their male counterparts.

Females in training are more than twice as likely to incur an injury during training and nearly four times as likely to experience an injury that results in hospitalization[3,28] or work time loss (i.e. stress fracture) than men[93]. Further, females in combat positions are up to ten times more likely to incur a musculoskeletal injury than their male counter parts, with up to a 100-fold higher risk increase of a pelvic stress fracture[49,50]. Anthropometric differences may lead to these increased instances of injury rates. On average, women have 30% and 50% less muscle mass in their lower body and upper body, respectively, than men. Additionally, female soldiers tend to be shorter, resulting in shorter natural stride lengths and increased stride frequency to maintain company pace[4,94]. However, stride frequency can only be increased to a point before over-striding occurs in an attempt to sustain pace. The biomechanical adaptations associated with over-striding often lead to pelvic stress fractures[95]. Generally speaking, females tend to enter training at a lower level of physical fitness, therefore partially explaining the injury discrepancy between male and female recruits[3,28,96,97]. While all of these variables may contribute to the higher rates of injury in female military personnel, there is a severe lack of information regarding the biomechanical differences between male and female personnel while donning soldier-relevant body borne loads.

Sex Dimorphism in Load Carriage Biomechanics

Limited research has examined the impact of body borne loads commonly worn in the military on the lower limb biomechanics of females. When female backpackers donned body borne loads relative to body weight, similar load carriage adaptations to that of male-only military load carriage studies were documented[55]. Simpson et al. found that load proportionally increases ground reaction force acting on the body and stance time[55], a finding documented in load carriage research dedicated to male military personnel[15]. Additionally, load resulted in an increase of mediolateral impulse, resulting in less stable gait patterns[55], which may exacerbate the unstable kinematic patterns of the unloaded female[70,72] and increase risk of musculoskeletal injury. While running with load, Xu et al. reported that the addition of body borne load resulted in significantly higher peak hip flexion moments in females, but failed to elicit any significant rotational adaptations to load. While these studies document the adaptations females make to body borne load, it fails to compare those adaptations to that of males to examine the influence gender on load carriage biomechanics.

When males and females don body borne load relative to their weight and march at a self-determined speed, there is a lack of significant differences between the sexes in gait mechanics[35,36]. While examining a standardized load of 22 kg at a predetermined marching speed, there were no lower limb kinetic or kinematic variables were significantly different between the sexes[35]. Krupenevich et al. speculated that the lack of biomechanical adaptation to the addition of load may lead to higher injury rates among female trainees and soldiers; If females fail to exhibit these sex-specific differences compared to males during load carriage, it may be that they are not making the adaptations necessary to safely carry standardized loads.

There is evidence that load carriage increases risk of injury by way of maladaptive biomechanics among soldiers on a large scale in both marching and running conditions. Additionally, there is ample research surrounding sex dimorphism present in both running, cutting and walking, and it is regularly reported that there are high instances of female musculoskeletal overuse injuries in the military service. With the recent addition of females to frontline and combat duties, there is a severe lack of understanding as to what underlying biomechanical factors may lend females to the high rates of injury in the service. The few studies that have examined sex dimorphism with a load have explored sex differences while marching. The field of military research would highly benefit from dissecting the implications of standardized load carriage on the sex dimorphism found while running.

CHAPTER THREE: MANUSCRIPT

Introduction

Musculoskeletal injuries are an extensive and ever-increasing problem for the military. In 2012, military personnel suffered 2.2 million musculoskeletal injuries with associated healthcare costs exceeding 700 million dollars[1]. These musculoskeletal injuries commonly occur during training[2] and are the most prominent cause of long-term disability and medical discharge for military peronnel[3]. A majority of these musculoskeletal injuries occur in the lower limb[4–6] when donning body borne load during military training-related tasks[10,11] Military body borne loads are composed of personal protective and fighting equipment and supplies[4] that routinely range from 20 to 40 kg[41]. The addition of these body borne loads during training-related tasks reportedly produces maladaptive lower limb biomechanical patterns that decrease physical performance[5] and increase the risk of sustaining a lower limb musculoskeletal injury[5,11].

Body borne load increases risk of suffering musculoskeletal injury during the performance of locomotor tasks common to military training, such as running and cutting [10,11,14]. During these locomotor tasks, the addition of body borne load results in a significant increase in the peak vertical ground reaction force acting on the musculoskeletal system[15–17]. To safely attenuate these elevated forces, military personnel increase stance time, decreasing locomotor speed[98], and alter lower limb joint angles and joint moments[18,19], increasing risk of injury. For instance, military personnel exhibit a significant reduction in peak hip and knee flexion angle when the body borne load is increased from 20 kg to 40 kg during military-relevant running and cutting tasks[14,18,20]. While these extended postures may prevent limb collapse, it aids the transfer of the elevated ground reaction forces to the limb[21], subsequently increasing the risk of bony[22,23] and/or soft-tissue injury[24]. Further increasing the risk of musculoskeletal injury during military related tasks are the elevated joint moments that are adopted with body borne load. During a single-leg cut with body borne load, Brown et al.[14] reported that military personnel exhibited a significant increase in peak hip and knee flexion, and hip adduction joint moments to successfully complete the maneuver. These elevated joint moments require greater force generation by the hip and knee musculature to prevent collapse of the limb, but increase the stresses placed on the musculoskeletal system and risk of injury[25,26,27]. Yet, to date, it is unknown if military personnel exhibit similar adaptations in lower limb biomechanics, and subsequent increased risk of musculoskeletal injury, when performing a single-leg cut with the small incremental changes of body borne load commonly worn during military related tasks (i.e. 20 to 35 kg).

During military training, female personnel are two and a half times more likely than their male counterparts to suffer a musculoskeletal injury[3,28]. This discrepancy in injury rate may be attributed to the sex dimorphism in lower limb biomechanics exhibited during locomotor activities routinely performed during military training, such as single-leg cutting. During unloaded cutting, females exhibit decreased peak hip and knee flexion joint angles[29,30], and increased hip adduction, and knee flexion and abduction joint moments compared to their male counterparts[31,32]. The female's extended limb may be necessary to prevent collapse of the leg, but leads to increased risk of musculoskeletal injury from greater peak proximal tibial shear and subsequent loading of the joint's soft-tissues[23,33,34]. Despite their increased risk of musculoskeletal injury, it is currently unknown if females exhibit a sex dimorphism in lower limb biomechanics when performing a single-leg cut with body borne load. Previously, neither Krupenevich et al.[35] or Silder et al.[36] reported no significant sex difference in lower limb biomechanics when walking with small body borne loads, despite the fact males and females exhibit significant differences in hip and knee biomechanics during unloaded walking[37–39]. When running with small body borne loads, Xu et al. reported that females exhibited increases in vertical ground reaction force, joint reaction forces and tibial stresses, increasing their risk of musculoskeletal injury[34]. However, it is unknown if males and females exhibit similar to lower limb biomechanical adaptations during locomotor tasks, such as a single-leg cut, with body borne loads commonly worn during military training.

Considering females were recently given the opportunity to fill infantry positions in the U.S. Military[41], where they are routinely required to run and cut with body borne load, it is imperative to quantify and compare the lower limb biomechanical adaptions they exhibit with male personnel during performance of locomotor tasks while donning the body borne loads common to military activities. With that in mind, the purpose of this study was to determine the lower limb biomechanical adaptations exhibited during a single-leg cut with body borne loads (20, 25, 30 and 35 kg) commonly worn during military activities, and compare whether the lower limb biomechanical adaptations as a result of the body borne load significantly differ between male and female participants. It is hypothesized that the addition of body borne load will result in reductions of hip and knee flexion, hip adduction angle, increases in the corresponding moments and elevated proximal tibial shear. Further, it is hypothesized that females will exhibit further reductions in hip and knee flexion, increased hip adduction and knee external rotation postures and moments, and increased proximal tibial shear.

Methods

Subjects

Twenty-eight (17 Male and 11 Female) participants were recruited for this study (Table 3.1). Each potential participant self-reported the ability to safely carry 75 pounds, but were excluded if they have: (1) a history of previous back or lower extremity injury or surgery, (2) pain in back or lower extremity prior to testing, (3) any recent injury to the back or lower extremity (previous 6 months), (4) any known neurological disorder, and (5) are currently pregnant. Research approval was acquired from the local Institutional Review Board and all participants provided written informed consent prior to testing.

Table 3.1 Average subject demographics (SD).

	N	Height (m)	Weight (kg)	Age (years)
Males	17	1.79 (0.08)	81.69 (9.42)	21.33 (2.77)
Females	11	1.66 (0.03)	66.86 (8.18)	21.92 (1.97)

Biomechanical Testing

All participants performed one orientation session and four test sessions. During the orientation session, subject strength data was collected using an isokinetic dynamometer (System 2, Biodex Medical Systems, Inc, Shirley, NY, USA). Each participant completed maximal-effort isometric hip flexion, extension, abduction and adduction, and knee flexion and extension contractions with their dominant limb. Hip flexion and extension was measured with a neutral joint position of 0 degrees[99], while hip adduction and abduction used a neutral joint position of 10 degrees[100]. Knee flexion and extension required a starting joint position of 45 and 60 degrees, respectively[101]. Participants were required to complete three repetitions of 3-second maximal effort muscular contractions with a 30-second rest period between each repetition. The maximum force production (in N) was recorded for each trial.

During each test session, participants performed the single-leg cutting task with a different body borne load (*20, 25, 30 and 35 kg*). For each load, participants were required to wear spandex shirt and shorts, weighted vest (Box ®, WeightVest.com, Inc., Rexburg, ID, USA), and standard-issue military helmet (ACH), and carry a mock weapon (M16) (Picture 1). The helmet and mock weapon weighed approximately 6.17 kg. To apply the additional load for each condition, participants donned a weighted vest, which was systematically adjusted to add the load required for each condition to the participant's torso. The total weight of each condition was required to be within ± 2 % of the targeted load (i.e., *20, 25, 30 or 35-kg*). To ensure each load met this requirement, participants were weighed at the start of each test session. To randomize and counter-balance the test order, a 4 x 4 Latin square was used to assign a sequence of load conditions to each participant, prior to beginning the study. Each test session was separated by a minimum of 24 hours to minimize the effects of fatigue and limit chances of injury.

Picture 1 Equipment (helmet, mock weapon and weighted vest) for each body borne load condition (20, 25, 30 and 35 kg).

During each test session, participants had 3D trunk and lower limb (hip, knee and ankle) biomechanical data recorded during a series of dynamic, single-cuts. During each cut, eight high-speed (240 fps) optical cameras (MXF20, Vicon Motion Systems, Ltd., London, UK) captured trunk and lower limb motion data, while two force platforms (AMTI OR6 Series, Advanced Mechanical Technology, Inc., Watertown, MA.) embedded in the laboratory floor recorded synchronous ground reaction force (GRF) data (2400 Hz). The single-leg cut required participants to run at 4.0 m/s ± 5 % through the motion capture volume before planting their foot on the force platform and performing a 45° cut towards the opposite limb. For the cut left, participants planted their right foot on the force platform and cut 45° towards the left; whereas for cut right, participants planted with their left foot on the force platform and cut 45° towards the right. The direction of each cut was randomized using the random number generator (Excel 2016, Microsoft, Seattle, WA, USA) prior to each test session. During each cut, running speed was quantified from two sets of infrared timing gates (TF100, ©TracTonix, Lenexa, KS, USA) placed 4 meters apart in the motion capture area immediately preceding the force platform. Each participant performed five successful cuts of each foot (right and left). A cut was considered successful if the participant: cut at the required angle (45° ± 5°), only contacted the force platform with the required foot, and ran the predetermined speed. During testing, participants were required to rest between trials to minimize the effects of fatigue.

Biomechanical Analyses

During each single-leg cut, trunk and lower limb biomechanics were quantified from the 3D trajectories of 34 retro-reflective and four virtual markers (Table 3.2). Each reflective marker was attached over a specific landmark with double-sided tape (Sensi-Tak Tape Roll, Walker Tape, West Jordan, UT, USA) and secured with elastic tape (Cover-Roll Stretch Tape, BSN Medical, Charlotte, NC, USA). Virtual markers were created on the torso (specifically, sternum jugular notch, xiphoid process, cervical vertebrae 7, and midpoint between the inferior angles of the scapulae) by digitizing their location in the global coordinate system using a Davis Digitizing Pointer (C-Motion, Inc, Germantown, MD, USA). After marker placement, each participant stood in anatomical position to have a stationary recording taken. The stationary recording was used to create a kinematic model in Visual 3D (v6, C-Motion, Inc, Germantown, MD, USA). The kinematic model consisted of eight rigid kinematic segments (pelvis, trunk and bilateral foot, shank and thigh) and had 27 degrees of freedom (Table 3.3). Each segment of the kinematic model had an orthogonal (local) coordinate system with an origin located at a virtual joint center and three orthogonal axes (x, y and z) that follow assigned according to the right hand principle in Visual 3D. The pelvis was defined with respect to the global (laboratory) coordinate system and assigned six (three rotation and three translational) degrees of freedom. For the hip, a functional joint center was calculated according to Rozumalski and Schwartz[102] and local coordinate system assigned three degrees of freedom. For the knee, the joint center was calculated as the midpoint between the lateral and medial femoral epicondyles, and local coordinate system assigned three degrees of freedom according to Grood and Suntay[103]. For the ankle, the joint center was calculated as the midpoint between the lateral and medial malleoli and local coordinate system assigned three degrees of freedom according to Wu et al.[104]. For the trunk, the origin was defined as the intersection between the midpoint of the acromion processes and the midpoint between the sternum jugular notch and seventh cervical vertebrae and assigned three degrees of freedom, according to Wu et al.[104].

Table 3.2 Placement of 34 retroreflective markers for the kinematic model.

	Markers
Trunk	*sternum jugular notch, xiphoid process, the seventh cervical vertebrae, and midpoint between the inferior angles of the scapulae,* **acromion process**
Pelvis	**posterior superior iliac spine**, **anterior superior iliac spine**, superior iliac crest
Thigh	greater trochanter, **medial and lateral femoral epicondyles**, thigh proximal to patella
Shank	tibial tuberosity, lateral fibula, distal tibia, **medial and lateral malleoli**
Foot	**first and fifth metatarsal heads**, heel, dorsal portion of foot between first and fifth metatarsal heads.

Note: *italics* denotes virtual markers and **bold** denotes calibration markers.

For each single-leg cut, synchronous GRF and 3D marker trajectories were low-pass filtered with a fourth-order Butterworth filter (12 Hz). The GRF and marker data were filtered with the same cut-off frequency, as it has been shown to improve accuracy of kinetic parameters[105]. The filtered marker trajectories were processed by Visual 3D to calculate 3D joint rotations of the lower limb. The joint rotations were expressed relative to each participant's anatomical position (stationary recording), using a joint coordinate systems approach[103,106].

Filtered kinematic and GRF data were processed using conventional inverse dynamic analysis to obtain 3D intersegmental forces and moments at each lower limb joint[107]. Inertial properties of each segment were defined according to Dempster[108]. Hip, knee and ankle 3D forces were transformed to respective distal (femoral, tibial and talar) segment reference frames and anterior–posterior, medial–lateral and compression–distraction forces were be calculated. The intersegmental moments were characterized with respect to the cardanic axes of their respective joint coordinate systems[107]. At the hip and knee, the intersegmental moments were defined as flexion-extension, abduction-adduction and internal-external rotation, while at the ankle the moments were defined as dorsiflexion-plantarflexion and inversion-eversion. Joint moments were normalized by body mass (kg) and height (m), and expressed as external moments. Forces were normalized by body weight (N) and positive direction was expressed according to the corresponding orthogonal axis (i.e., peak proximal tibial shear will be defined as the anteriorly directed y-axis force on the proximal tibia). All biomechanical data was time-normalized to 100 % of stance phase and resampled at 1% increments (N = 101). Stance phase (0% - 100%) was defined as heel strike to toe off. With heel strike and toe off occurring at the first instant the GRF exceeds and falls below 10 N, respectively.

Statistical Analysis

Biomechanical variables reported to be related to risk of musculoskeletal injury were selected for statistical comparison. Specifically, the kinematic dependent variables are initial contact (IC) and peak of stance (PS; 0% - 100% of stance phase) hip flexion, adduction and internal rotation, and knee flexion, abduction and external rotation, and ankle dorsiflexion and eversion. The kinetic dependent variables are PS hip flexion, adduction and internal rotation, and knee flexion, abduction and external rotation, and ankle dorsiflexion and eversion joint moments, and peak proximal tibial shear force. Each dependent variable was averaged across three successful trials to create a participant-based mean. Additionally, coefficient of variation (CV) was calculated to measure variation of each variable of interests between trials for each participant[109]. Then, the participant-based means and CV were submitted to a repeated measures ANOVA to test the main effects of and interaction between sex (*male*, *female*), load (*20, 25, 30 and 35kg*) and limb (*dominant, non-dominant)*. Where statistically significant ($p \leq 0.05$) differences are observed, a Bonferroni procedure[110,111] was used. For each pairwise comparison, effect size was calculated using Cohen's d[112]. Independent t-tests were used to compare weight, height and hip and knee strength measures between sexes. All statistical analysis was completed using SPSS (v23, IBM Corporation, Armonk, New York, USA). Alpha level will be set *a priori* at $P < 0.05$.

Results

Joint Angles

No significant interactions were observed for joint angles and thus, only main effects are presented for IC and PS joint angles. Body borne load had no significant effect on any IC or PS hip, knee or ankle joint angle. Females exhibited significantly greater IC ($p = 0.016$, $d = 1.028$) and PS ($p = 0.015$, $d = 0.999$) hip adduction angle compared to males (Figure 3.1B; Table C.1-2). But, sex had no significant effect on any other IC or PS hip, knee or ankle angle. The non-dominant limb exhibited significantly greater IC ($p=0.015$, $d = 0.419$) and PS ($p = 0.05$, $d = 0.474$) hip internal rotation angle compared to the dominant limb (Figure 3.2C). Limb dominance, however, did not have a significant effect on any other IC or PS hip, knee or ankle angle.

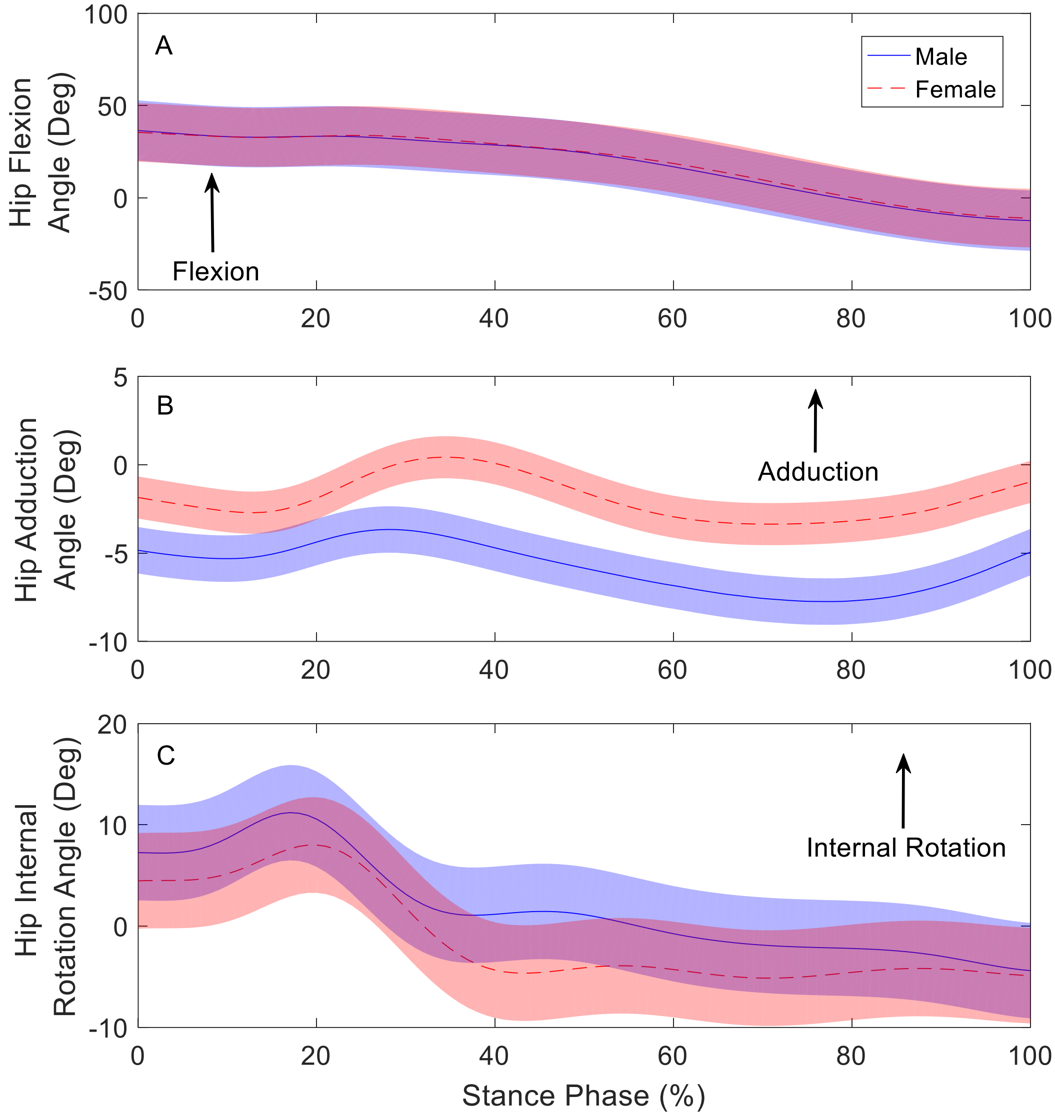

Figure 3.1 Stance phase (0% - 100%) hip flexion (A), adduction (B) and internal rotation (C) joint angle for the male and female participants during the single-leg cut.

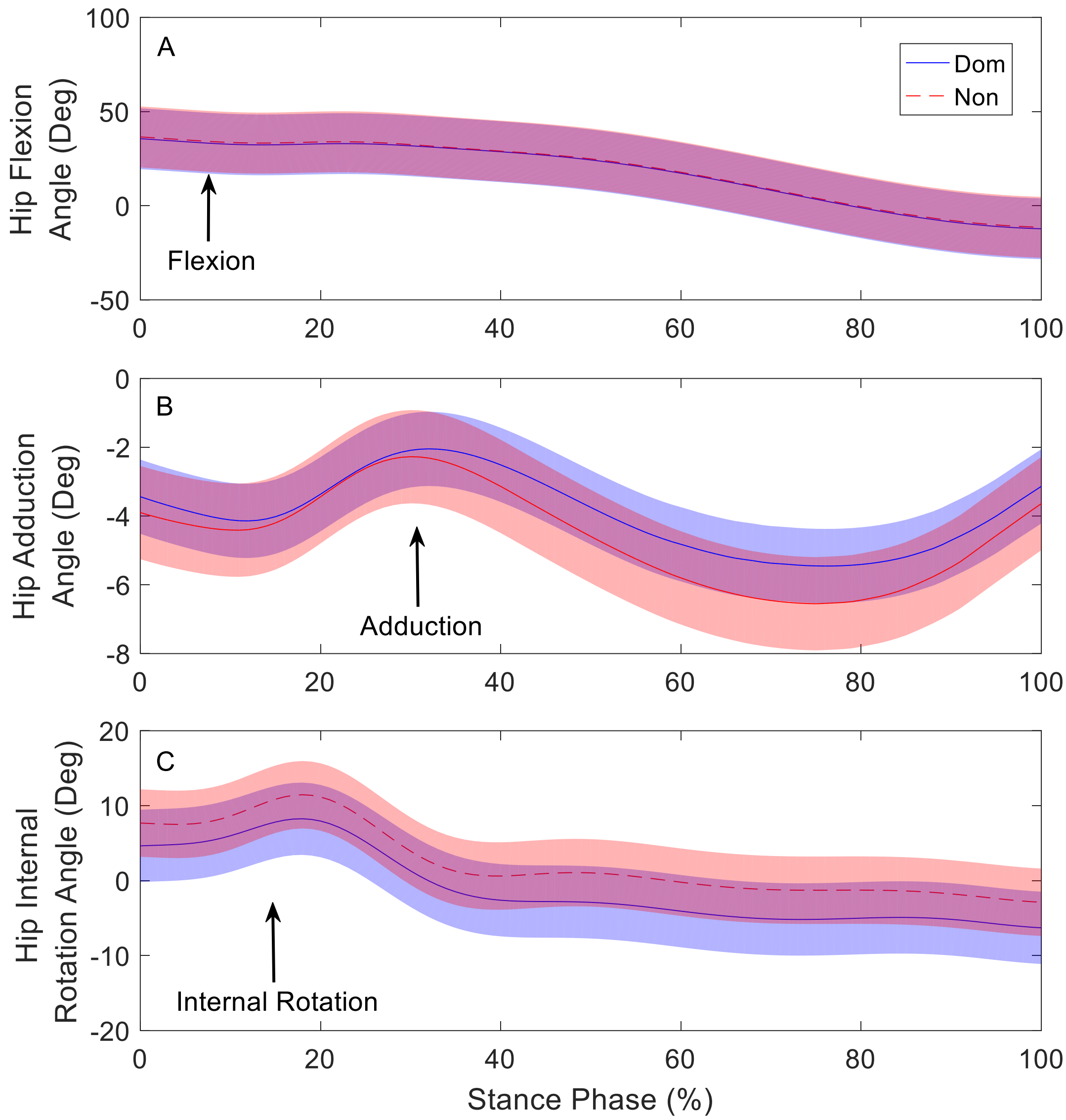

Figure 3.2 Stance phase (0% - 100%) hip flexion (A), adduction (B) and internal rotation (C) joint angle for the dominant and non-dominant limb during the single-leg cut.

Joint Moments

A significant limb by sex interaction was observed for PS hip internal rotation moment (Figure 3.3C; p = 0.004). Females exhibited significantly greater hip internal rotation moment compared to males with their dominant (p = 0.017, *d* = 0.927), but not with in their non-dominant limb (p = 0.243). But, males exhibited significantly greater hip internal rotation moment with their non-dominant limb compared to dominant limb (Figure 3.3A; $p < 0.001$, $d = 1.556$), while females had no significant difference between limbs (Figure 3.3B; $p = 0.873$).

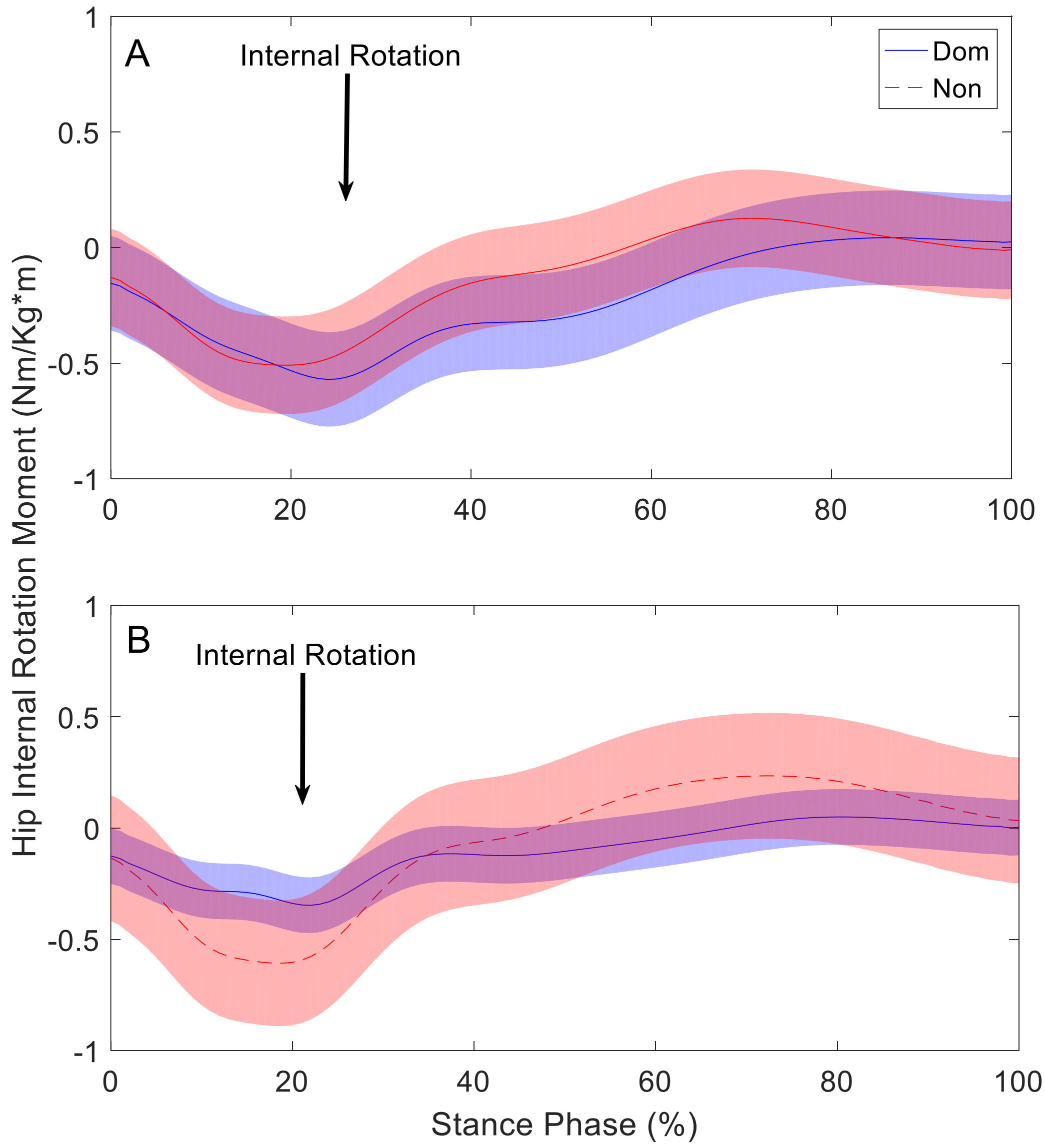

Figure 3.3 Stance phase (0% - 100%) hip internal rotation moment for males (A) and females (B) for both limbs during the single-leg cut.

There was a significant load by sex interaction for PS knee flexion moment (Table C.3; $p = 0.015$). Males exhibited significantly greater PS knee flexion moment compared to females with the 25 kg ($p = 0.040$, $d = 0.864$) and 30 kg ($p = 0.022$, $d = 0.999$) loads, but no sex differences were observed between the 20 kg ($p = 0.920$) and 35 kg ($p = 0.119$) loads. Additionally, males exhibited a significant increase in PS knee flexion moment with the 25 kg compared to 20 kg ($p = 0.018$, $d = 0.644$) load, but no significant difference was present between any other loads.

There is a significant load by limb interaction (Table C.3) for PS knee abduction moment ($p = 0.024$). The non-dominant limb exhibited greater PS knee abduction moment with the 35 kg compared to 20 kg ($p = 0.003$, $d = 0.589$) load, but no significant difference between any other load. The non-dominant limb also exhibited significantly greater PS knee abduction moment compared to the dominant limb with 25 kg ($p = 0.047$, $d = 0.476$), but not for the 20 kg ($p = 0.314$), 30 kg ($p = 0.641$) or 35 kg ($p = 0.055$) loads (Table C.3).

Body borne load only had a significant effect on PS hip flexion moment (Figure 3.4A; $p = 0.026$), but not any other PS hip, knee or ankle joint moment. Specifically, PS hip flexion moment was significantly greater with the 35 kg compared to 20 kg load ($p = 0.017$, $d = 0.486$), but no significant differences were evident between any other load.

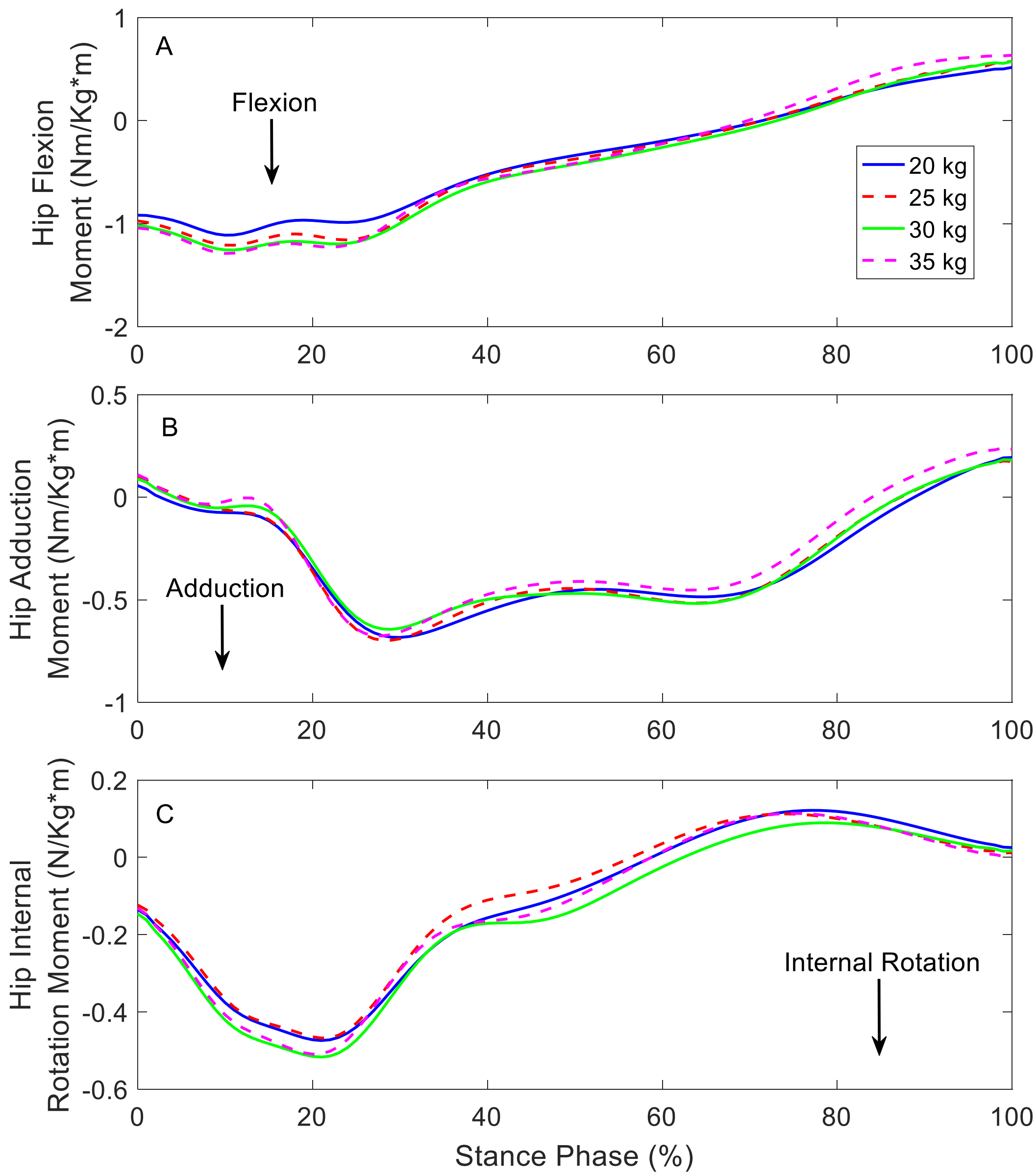

Figure 3.4 Stance phase (0% - 100%) hip flexion (A), adduction (B) and internal rotation (C) moment during the single-leg cut for each load configuration (20, 25, 30 and 35 kg).

During the cut, males exhibited significantly larger PS hip flexion moment compared to females (Figure 3.5A; p = 0.041, *d* = 0.873), while females exhibited significantly greater PS hip adduction (Figure 3.5B; p < 0.001, *d* = 1.727) and knee external rotation (Figure 3.6C; p = 0.004, *d* = 1.249) moments compared to males. But, sex had no significant effect on any other PS hip, knee or ankle moment.

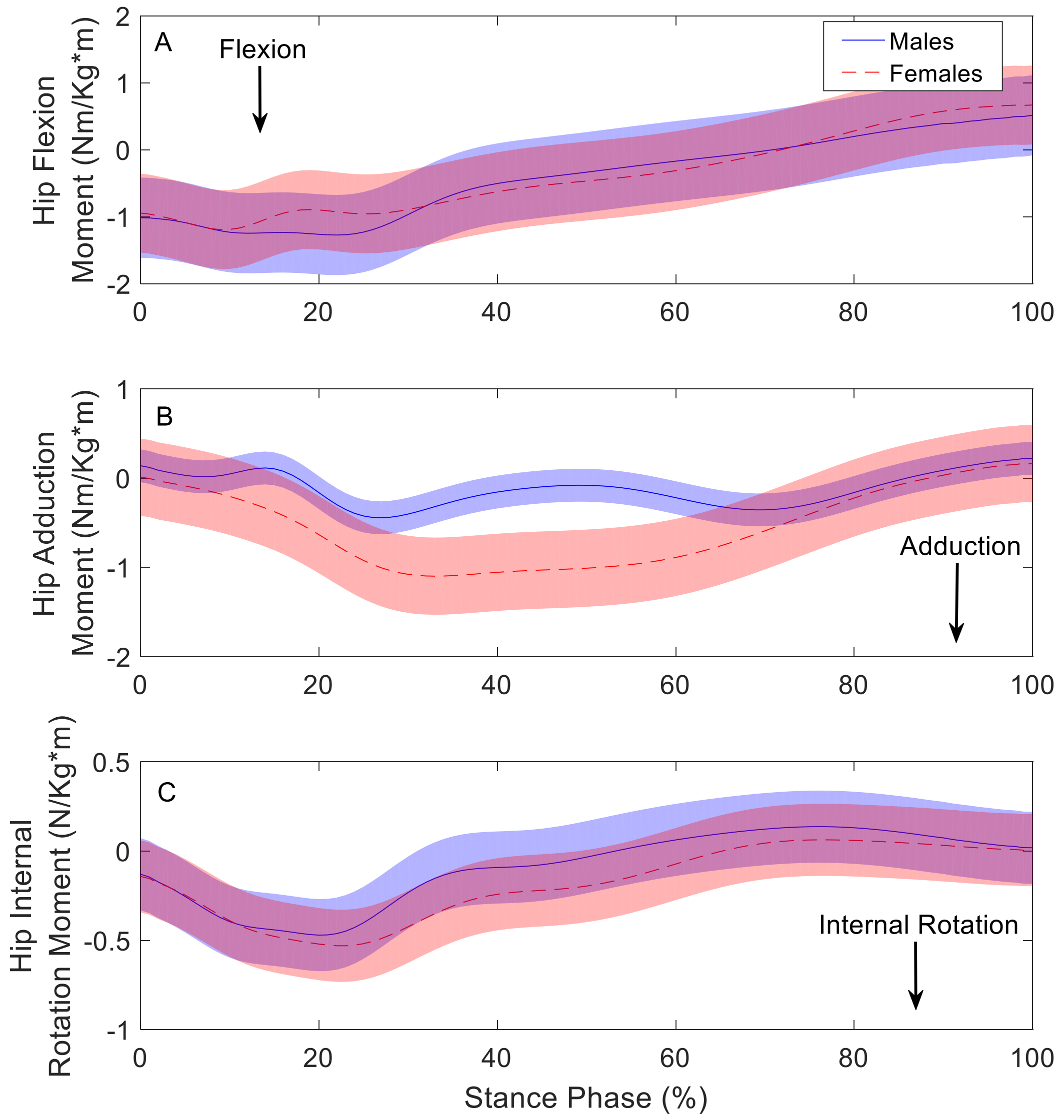

Figure 3.5 Stance phase (0% - 100%) hip flexion (A), adduction (B) and internal rotation (C) moment during the single-leg cut for males and females.

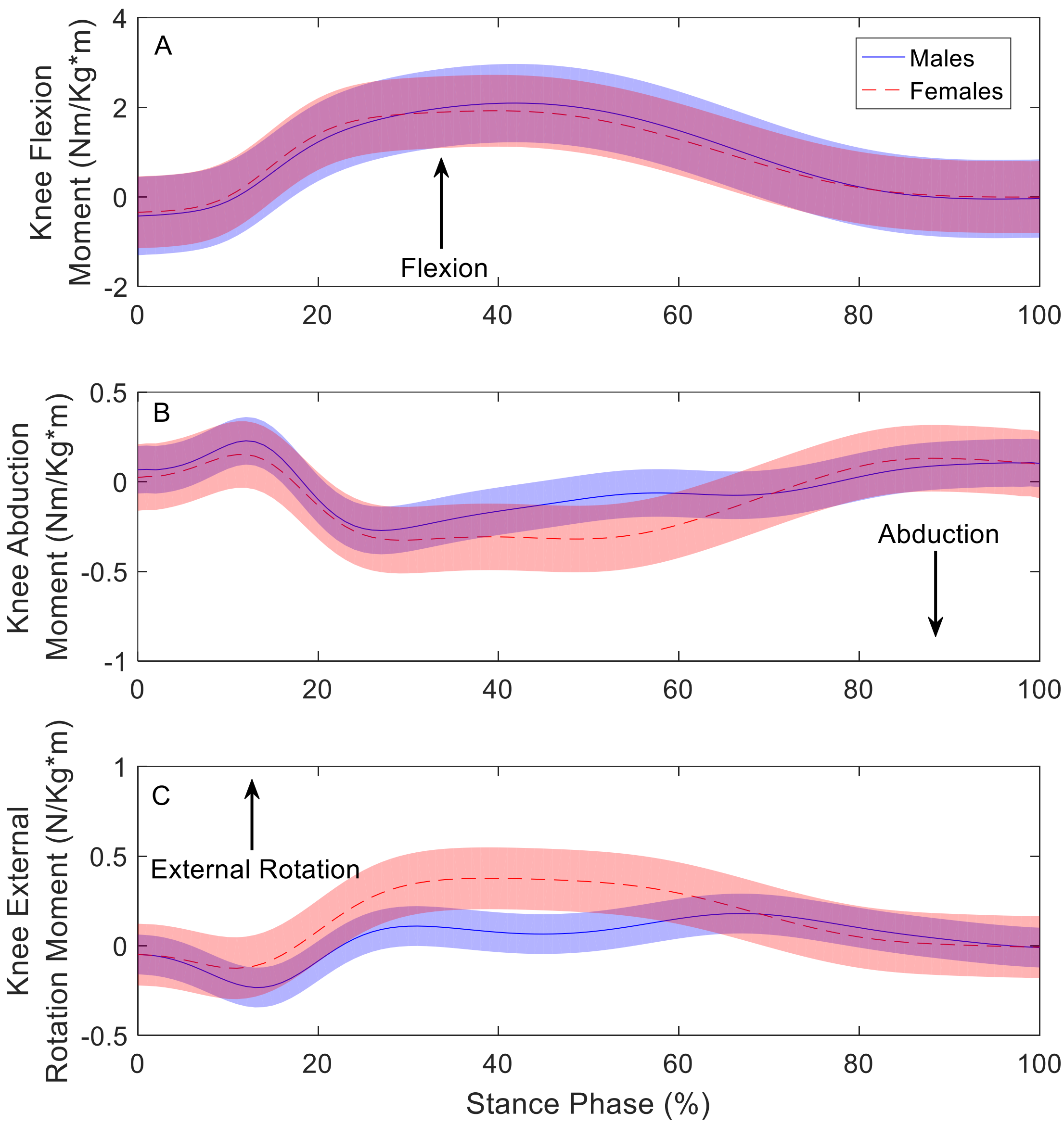

Figure 3.6 Stance phase (0% - 100%) knee flexion (A), abduction (B) and external rotation (C) moment during the single-leg cut for males and females.

The dominant limb exhibited significantly greater PS hip adduction moment (Figure 3.7B; p = 0.010, *d* = 0.438) compared to the non-dominant limb, whereas, the non-dominant limb exhibited significantly greater PS hip internal rotation (Figure 3.7C; p = 0.007, *d* = 0.627) and knee external rotation (Figure 3.8C; p = 0.003, *d* = 0.821) moments compared to dominant limb. Limb had no significant effect any other peak hip, knee or ankle peak moments.

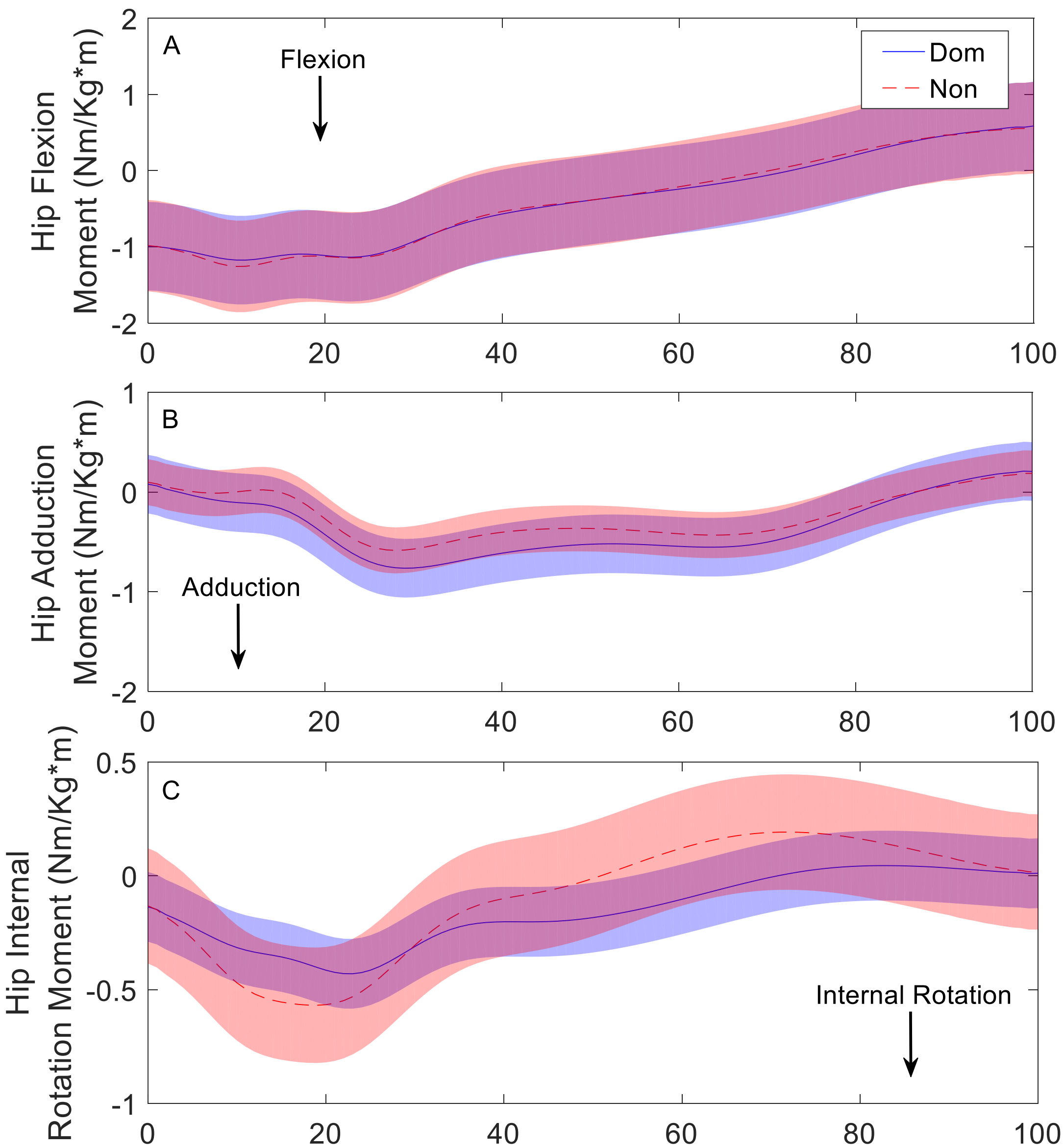

Figure 3.7 Stance phase (0% - 100%) hip flexion (A), adduction (B) and internal rotation (C) moment during the single-leg cut for dominant and non-dominant limbs.

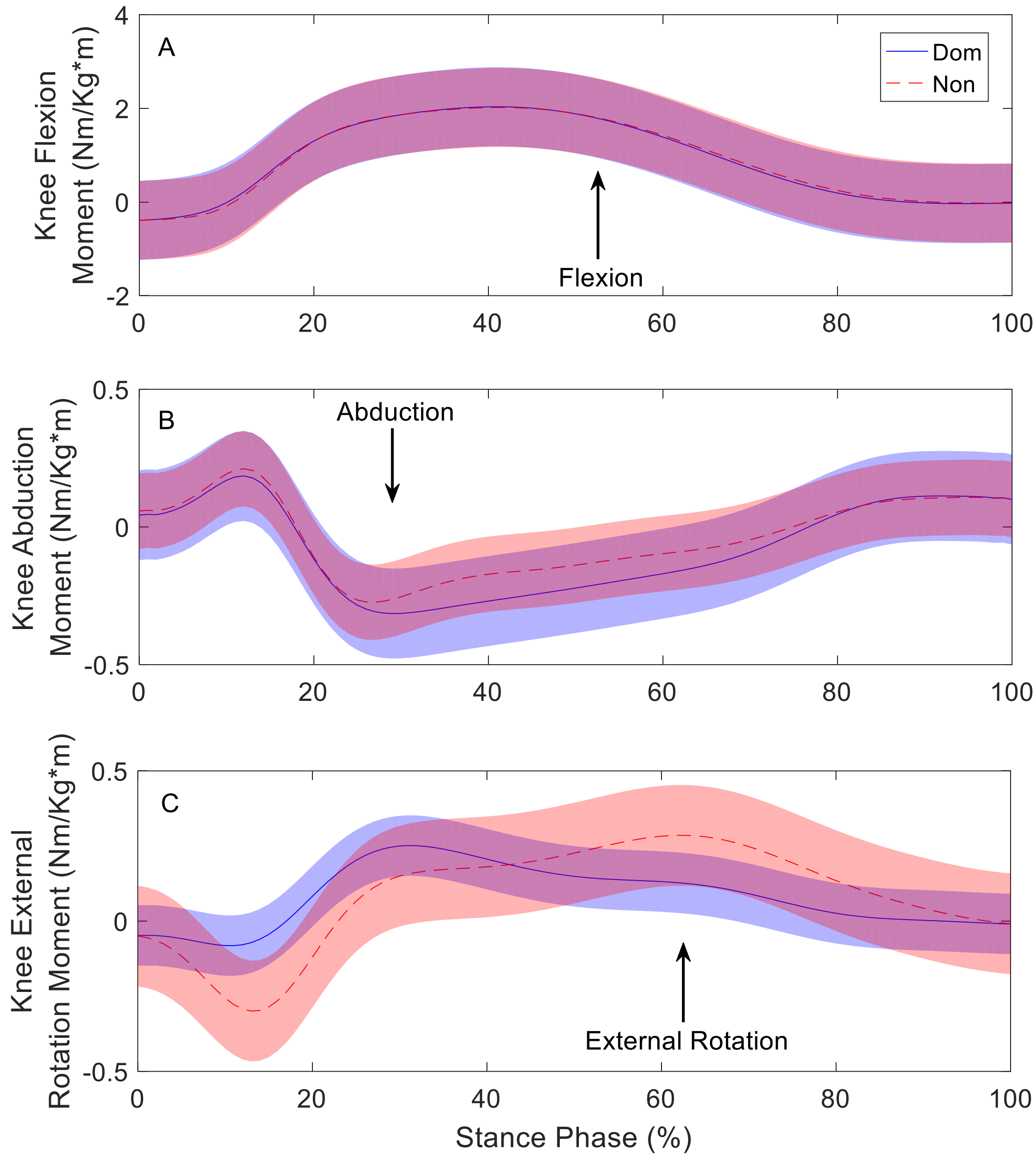

Figure 3.8 Stance phase (0% - 100%) knee flexion (A), abduction (B) and external rotation (C) moment during the single-leg cut for dominant and non-dominant limbs.

<u>Proximal Tibial Shear</u>

There was a significant sex by load interaction for peak proximal tibial shear (Table 3.4; p = 0.037). Females exhibited significantly greater peak proximal tibial shear with the 25 kg compared to 20 kg load (p = 0.028, *d* = 0.578), but significant differences were not observed between any other loads. Males, however, exhibited significantly greater proximal tibial shear with the 35 kg compared to 20 kg (p = 0.040, *d* = 0.351) and

25 kg loads (p = 0.011, *d* = 0.381), and not with the 30 kg load (p = 0.066). The proximal tibial shear exhibited by males in the 25 kg compared to the 20 kg condition was not significant (p = 1.000), as the values were identical.

Body borne load had a significant effect on peak proximal tibial shear (p = 0.004). Peak proximal tibial shear was significantly larger with the 35 kg compared to 20 kg (p = 0.011, *d* = 0.370) load. But, no significant differences were observed between any of the other loads. Neither sex (p = 0.298) or limb (p = 0.633) had a significant effect on peak proximal tibial shear.

Table 3.3 Mean (SD) proximal anterior tibial shear (Wt) in the dominant and non-dominant limbs for males and females.

	20 kg		**25 kg**		**30 kg**		**35 kg**	
	Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
M	0.105 (0.065)	0.103 (0.069)	0.106 (0.060)	0.101 (0.060)	0.110 (0.056)	0.010 (0.064)	0.131 (0.063)	0.121 (0.066)
F	0.120 (0.044)	0.110 (0.049)	0.130 (0.039)	0.151 (0.057)	0.133 (0.039)	0.133 (0.043)	0.135 (0.047)	0.134 (0.054)

Subject Demographics

Males were significantly taller (p < 0.001, *d* = 1.78), heavier (p = 0.001, *d* = 1.43) and stronger than their female counterparts (Table 3.1, Table 3.5). Specifically, males exhibited greater hip and knee flexion (p = 0.028, *d* = 0.968 and p =0.009, *d* = 0.171) and extension (p = 0.040, *d* = 0.921 and p = 0.011, *d* = 1.095), and hip abduction (p = 0.013, *d* = 0.516) strength than female participants. No effect of sex was observed for hip adduction (p = 0.916) strength.

Table 3.4 Mean (SD) subject maximum hip and knee strength measures (%BW) for male and female participants.

Hip Flx*	Hip Ext*	Hip Add	Hip Abd*	Knee Flx*	Knee Ext*
64.06 (24.22)	64.63 (27.29)	46.89 (10.10)	45.16 (11.44)	48.76 (10.15)	71.34 (13.54)
45.81 (11.15)	46.45 (5.83)	47.35 (13.66)	32.91 (12.42)	38.25 (8.62)	57.59 (11.49)

*Denotes a significant main effect ($p < 0.05$) of sex.

<u>Coefficient of Variation</u>

<u>Joint Rotations</u>

No significant interactions were observed for CV of IC and PS joint angles and thus, only main effects are presented. The dominant limb exhibited greater variation of IC and PS hip flexion (Tables C.4-5: p = 0.030, *d* = 0.53 and p = 0.007, *d* = 0.51) angle than the non-dominant limb. But, the non-dominant limb exhibited more variation of IC knee external rotation angle (p = 0.049, *d* = 0.42) than in the dominant limb. Limb had no significant effect on any other IC or PS joint angle, and neither sex or load impacted the variation of any joint angle.

<u>Joint Moments</u>

A significant limb and sex interaction (p = 0.002) was observed for variation of PS hip internal rotation moment. Males exhibited greater variation of hip internal rotation moment than females in the dominant limb (p = 0.034, *d* = 0.92); whereas, females exhibited greater variation in the non-dominant limb (p = 0.027, *d* = 0.87) than males. Females exhibited more variable hip internal rotation moment in non-dominant limb compared to their dominant limb (p < 0.001, *d* = 1.02), while males exhibited no differences between limbs (p = 0.118).

The non-dominant limb exhibited greater variation for hip internal rotation ($p < 0.001$, $d = 1.14$) and knee external rotation ($p = 0.045$, $d = 0.50$) moment than the dominant limb. Additionally, males exhibited significantly more variability hip adduction moment ($p = 0.017$, $d = 1.09$) than their female counterparts. Limb or sex did not influence any other joint moments, and load had no effect of variation of any joint moment.

Proximal Anterior Tibial Shear

Sex, load or limb on no effect on the variation of proximal anterior tibial shear force.

Discussion

Training-related musculoskeletal injuries are a common problem in the military[113]. These musculoskeletal injuries are thought to be a consequence of maladaptive lower limb biomechanics exhibited by military personnel performing training-related locomotor tasks, such as a single-leg cut. Brown et al. previously reported military personnel exhibit significant adaptations of hip and knee biomechanics thought to increase injury risk and decrease physical performance with the addition of 20 kg to body borne load[14]. This study tested the hypothesis that small, incremental increases of body borne load (5 kg, 10 kg and 15 kg) would result in significant adaptations in hip, knee and ankle joint angles and moments, but the current outcomes only partially support this hypothesis. In contrast to previous work[14,18], current participants did not exhibit a significant adaptation in IC or PS hip, knee or ankle joint angles when adding 5 kg, 10 kg or 15 kg of body borne load during the single-leg cut. While the reason for the current discrepancy in findings is not immediately evident, it may be that the small, incremental increases in the chosen body borne loads are not sufficient to burden the lower limb musculature and require a significant alteration of joint angles when performing the single-leg cut. Yet, despite not exhibiting adaptations of hip, knee and ankle joint angles with the addition of load, participants exhibited a significant increase in hip joint moments and knee joint forces when performing loaded single-leg cuts. While these adaptations may be necessary to successfully execute the single-leg cut task, they may also increase risk of knee musculoskeletal injury and decrease performance of the cutting task with body borne load.

Adding body borne load may increase risk of suffering a knee musculoskeletal injury. During the single-leg cut, participants exhibited a significant increase in peak proximal anterior tibial shear with the 15 kg addition of body borne load. Large magnitudes of an anteriorly directed proximal tibial shear force reportedly places greater loads on the knee's soft-tissue structures that restrain anterior translation of the tibia (i.e. the anterior cruciate ligament (ACL))[49,114]. Consequently, this shear force may indicate loading of the ACL[115] and could be considered a risk factor for injury. The addition of body borne load may further increase female's adoption of these "hazardous" shear forces at the knee and subsequent risk of suffering a soft tissue injury compared their male counterparts[116]. In agreement with Chappell et al.[117] females exhibited 20% larger peak proximal tibial anterior shear compared to their male counterparts[118]. Further, females exhibited a significant increase in proximal tibial shear force with only the 5 kg addition of body borne load; whereas, males did not exhibit a significant increase in proximal tibial shear force until there was a 15 kg difference in body borne load configurations. During the performance of dynamic locomotor tasks, such as the single-leg cut, females reportedly exhibit a quadriceps-dominant neuromuscular strategy[91]. This neuromuscular strategy is characterized by a heavy reliance upon and increased recruitment of the knee extensor muscles to successfully complete dynamic locomotor tasks[89]. Considering a significant increase in quadriceps activity is exhibited during load carriage[36] and purported to increase proximal tibial shear force[119], it may be the underlying cause of the elevated “hazardous” shear forces observed at the knee, particularly for females who must compensate for their weaker knee extensors. However, further research is warranted to determine if a quadriceps-dominant neuromuscular strategy further increases the risk of soft tissue injury at the knee during loaded single-leg cuts, and whether female military personnel are more reliant upon a quadriceps-dominant neuromuscular strategy during load carriage than their male counterparts.

To complete the single-leg cut, females exhibited hip and knee biomechanics related to dynamic valgus loading of the knee. In agreement with previous literature[76,82], there was a sex dimorphism in frontal and transverse plane hip and knee biomechanics during the single-leg cut. Specifically, females exhibited significantly greater hip adduction angle and moment, and hip internal rotation and knee external rotation moments compared to males during the single-leg cut. These hip and knee biomechanics may increase female’s risk of suffering a musculoskeletal injury by producing a subsequent increase in valgus loading at the knee[120]. Knee valgus loads are thought to strain the structures associated with stabilization of the joint[74], and may increase risk of the injury. Future research is warranted to examine if this increased knee valgus loading in female military personnel is present in other military-related tasks, and if body borne load further increases valgus loading at the knee during these movements.

Body borne load may also increase the risk of musculoskeletal injury for the non-dominant limb. Although the previous literature is inconclusive on whether substantial differences in lower limb biomechanics exist between the dominant and non-dominant limb[121,122], the current outcomes support significant limb dimorphism in lower limb biomechanics during the single-leg cut. During the single-leg cut, both the dominant and non-dominant limb exhibited hip and knee biomechanics associated with dynamic valgus. But, the non-dominant limb exhibited a significant increase in biomechanical patterns reported to be a predictor of soft-tissue injury at the knee joint[123] with the addition of load. During the single-leg cut, the dominant limb exhibited significantly greater peak hip adduction moment compared to the non-dominant limb. The hip adductors of the dominant limb are reportedly stronger than those of the non-dominant limb[124], and may have afforded the stronger, dominant limb greater use of that musculature to successfully execute the single-leg cut. Whereas, the reportedly weaker non-dominant limb exhibited greater and more variable hip and knee biomechanics, including greater hip internal rotation and knee abduction moments, and more variable knee external rotation angle and moment compared to the dominant limb. These biomechanics are thought to increase valgus loading of the knee and may subsequently risk of injury in the non-dominant limb[49,125,126]. The addition of body borne load may further elevate the risk of injury for the non-dominant limb, but not dominant limb. The non-dominant leg exhibited a significant increase in peak knee abduction moment between 20% and 40% when adding 5 kg through 15 kg of body borne load. Considering large knee abduction moments are a prospective predictor for ACL injury[27], the increase in knee abduction moment currently observed with the addition of body borne load for the non-dominant limb may also elevate its risk of injury. Future work is needed to determine whether the incidence of musculoskeletal injury differs between the dominant and non-dominant limb during military training, and whether training can reduce the non-dominant limb's elevated risk of injury.

To successfully execute the single-leg cut, the large, proximal hip musculature must power forward propulsion[127] and stabilize the trunk[128] during the change of direction. With the addition of the current torso borne loads, participants exhibited a significant increase of peak hip flexion moment to complete the cut. The heavier trunk required participants increase contraction of the large hip muscles, which increases strain on the soft tissue structures of the joint and decreases their ability to perform the maneuver[14].

To complete the single-leg cut, males exhibited larger sagittal plane hip and knee moments. In agreement with Pollard et al.[82] and Sigward et al.[91], males exhibited significantly greater hip and knee flexion moments compared to females during a single-leg cut. In agreement with previous literature, the significantly taller and heavier male participants exhibited stronger hip and knee musculature[129] than females. The increased lower limb strength may allow males to rely upon large muscles that produce hip and knee sagittal plane moments to successfully execute the cut task. While increasing hip and knee flexion moments may have afforded males successful performance of the single-leg cut, it may increase strain on the musculoskeletal system by aiding in the transfer of high ground reaction forces[21] to the soft tissue structures of the limb. Alternatively, females may not possess the strength of the lower limb musculature to adequately control the hip and knee with flexion moments and thus, adopt the use of frontal and transverse plane motions and loads to complete the single-leg cut[130,131].

The chosen male and female participants may be a limitation of the study. All participants self-reported the ability to safely carry body borne loads up to 75 pounds, but were not required to have significant load carriage experience. Participants who routinely carry body borne loads, such as military personnel, may exhibit a substantial difference in lower limb strength, leading to differences in biomechanics compared to inexperienced and/or weaker load carriers. Although, we are currently unaware of a significant difference in lower limb biomechanics exhibited by experienced and inexperienced load carriers, future study is warranted to determine whether load carriage experience and lower limb strength, rather than sex, impacts hip and knee joint moments during loaded cuts, and whether improving lower limb strength can reduce in the use of hip and knee biomechanics thought to increase risk of injury.

Conclusion

In conclusion, the addition of body borne load during execution of a single-leg cut lead to maladaptive lower limb biomechanics that may increase risk of knee injury in military personnel. With the addition of load, participants exhibited a significant increase of the anteriorly directed peak proximal tibial shear force, which is thought to strain the soft-tissue structures of the knee and increase injury risk. Additionally, both females and the non-dominant limb exhibited a dimorphism in lower limb biomechanics compared to males and the dominant limb, further elevating their risk of suffering a musculoskeletal injury during the single-leg cut. Compared to their stronger male counterparts, females produced a larger peak proximal tibial shear force, and frontal and transverse plane hip and knee biomechanics thought to contribute to valgus knee loads to complete the cut. During the single-leg cut, both limbs exhibited lower limb biomechanics that may contribute to valgus loading at the knee, but the addition of body borne load only produced a significant increase in knee abduction moment, and risk of soft-tissue injury, in the non-dominant limb.

CHAPTER FOUR: CONCLUSION

Introduction

This study's purpose was two-fold, to examine the influence of small incremental increases of body borne load on lower limb biomechanics exhibited during a single-leg cut, and to determine if men and women exhibit similar adaptations of lower limb biomechanics with the addition of body borne load during the single-leg cut. Key findings support the hypotheses that the addition of body borne load produces significant adaptations of lower limb biomechanics that increase risk of musculoskeletal injury, and females exhibit greater risk of musculoskeletal injury during the single-leg cut than their male counterparts.

Key Findings

Body borne load produced significant adaptation of lower limb biomechanics during the single-leg cut. Specifically, there was a main effect of body borne load for anteriorly directed peak proximal tibial shear force, peak vertical ground reaction force, the magnitude of the frontal and sagittal plane ground reaction force resultant vectors, and peak stance hip flexion moment. No effect of body borne load was observed on peak stance hip, knee or ankle joint angles during the single-leg cut.

There was a significant sex dimorphism in lower limb biomechanics during the loaded single-leg cut. Females exhibited greater anteriorly directed peak proximal tibial shear, hip adduction angles and moments, and knee external rotation moments compared to their male counterparts. Further, the addition of body borne load produced a significantly different adaption in the lower limb biomechanics for the male and female participants. Males experienced significantly higher peak knee flexion moments with the addition of load, though the same adaptation was not observed in females.

Significance

These findings are evidence that small incremental increases of body borne load significantly increase risk of musculoskeletal injury in military personnel. Further, this research is the first to show that the documented sex-dimorphism in an unloaded single-leg cut task is also present with the addition of body borne load, and that males and females differ in lower limb adaptations made in an effort to accommodate load, lending female military personnel to a higher risk of injury. The current findings are significant as this information can be used by the military to tailor the design of load carriage requirements to minimize injury risks associated with completing soldier-relevant tasks with the addition of load. Additionally, the military can use these findings to implement injury prevention and strength training protocol for military personnel to minimize musculoskeletal injury rates, particularly in females. By adequately preparing military personnel, specifically females, for the addition of body borne load during dynamic tasks, the adoption of hazardous lower limb biomechanics can be minimized and healthcare costs associated with musculoskeletal injuries can be lowered.

Limitations

The chosen participants may be a limitation of the study. Current participants were not required to have any load carriage experience, and as unexperienced load bearers, their adaptations to load may vary from those exhibited by personnel that are routinely required to carry body borne loads. Though all participants self-reported the ability to safely carry up to 75 pounds, strength of the current participants may be a limitation, as there were no other baseline strength or physical activity requirements. Additionally, the nature of combat positions requires that military personnel complete unanticipated cut tasks with the addition of load carried as a pack strapped to the back. For this reason, the use of strictly anticipated single-leg cutting maneuver while donning a weighted vest that evenly distributes the load around the torso is another limitation to this study.

Future Work

Future research should be dedicated to understanding the adaptations of lower limb biomechanics in those that are experienced in donning various loads often carried by military personnel in military-specific tasks. Further, additional research is warranted on the influence of similar increases of body borne load on other military-specific tasks, such as drop landing and running. Studying whether or not the implementation of a strength-training program can minimize dangerous lower limb adaptations to body borne load during the execution of military-specific tasks should also be a goal of future research. Limited research has been dedicated to understanding the influence of body borne load during unanticipated cuts, as such, further work is warranted to decipher how the addition of small, incremental increases of body borne load influence an unanticipated cut task and if the documented sex dimorphism is present in such a task.

As a neuromuscular sex dimorphism has been documented in dynamic, athletic tasks, future research should examine if females rely on the use of a quadriceps-dominant neuromuscular activation pattern to successfully complete a single-leg cut with the addition of body borne load compared to their male counterparts. Forthcoming research should also examine knee valgus loading in female personnel in other military-specific tasks to document if the addition of load results in a similar increase of knee valgus loading observed in the current study. Further, examining the influence of strength training protocol on female military personnel in an effort to reduce reliance on transverse and frontal plane motions and moments, the quadriceps-dominant neuromuscular pattern and subsequent dangerous proximal tibial shear force and valgus loading at the knee would be beneficial to the field.

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APPENDIX A

Pre-Participation Questionnaire

1. Have you suffered an injury to your hip, knee, or ankle in the past 6 months?

YES NO

If yes, please describe: ___________________________

2. Have you undergone surgery to your hip, knee, or ankle?

YES NO

If yes, please describe: ___________________________

3. Are you currently undergoing rigorous physical training or do you plan to start a rigorous training program in the next 3 months?

YES NO

If yes, please describe: ___________________________

4. Are you currently experiencing knee pain?

YES NO

5. Are you currently suffering from or have you ever suffered from a heart condition?

YES NO

If yes, please describe: ___________________________

6. Do you know of any reason why you cannot participate in this study?

YES NO

If yes, please explain: ____

I certify that the information I provided above is accurate.

Subject's Signature: ___________________________ Date: ______________

Subject's Name (Print): _________________________

Parent/Legal Guardian Signature: ___________________ Date: ___________

Parent/Legal Guardian Name (Print): _________________________

APPENDIX B

Footedness Questionnaire

Instructions: Answer each of the following questions as best you can. If you always use one foot to perform the described activity, circle **Ra** or **La** (for right always or left always). If you usually use one foot circle **Ru** or **Lu**, as appropriate. If you use both feet equally often, circle **Eq**.

Please do not simply circle one answer for all questions, but imagine yourself performing each activity in turn, and then mark the appropriate answer. If necessary, stop and pantomime the activity.

1. Which foot would you use to kick a stationary ball at a target straight in front of you?

La **Lu** **Eq** **Ru** **Ra**

2. If you had to stand on one foot, which foot would it be?

La **Lu** **Eq** **Ru** **Ra**

3. Which foot would you use to smooth sand at the beach?

La **Lu** **Eq** **Ru** **Ra**

4. If you had to step up onto a chair, which foot would you place on the chair first?

La **Lu** **Eq** **Ru** **Ra**

5. Which foot would you use to stomp on a fast-moving bug?

La **Lu** **Eq** **Ru** **Ra**

6. If you were to balance on one foot on a railway track, which foot would you use?

La **Lu** **Eq** **Ru** **Ra**

7. If you wanted to pick up a marble with your toes, which foot would you use?

La **Lu** **Eq** **Ru** **Ra**

8. If you had to hop on one foot, which foot would you use?

La **Lu** **Eq** **Ru** **Ra**

9. Which foot would you use to help push a shovel into the ground?

La **Lu** **Eq** **Ru** **Ra**

10. During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first?

La **Lu** **Eq** **Ru** **Ra**

11. Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities?

Yes **No**

12. Have you ever been given special training or encouragement to use a particular foot for certain activities?

Yes **No**

13. If you have answered YES for either question 11 or 12, please explain:

APPENDIX C

Table C.1 Mean (SD) Initial Contact hip, knee and ankle joint rotations (in degrees).

		20 kg		25 kg		30 kg		35 kg	
		Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
Hip	**M**	35.46(7.59)	36.08(6.49)	36.13(7.07)	37.51(6.36)	35.15(6.67)	36.40(5.73)	36.66(7.12)	38.46(7.01)
X	**F**	34.21(5.40)	35.35(4.44)	37.51(3.22)	36.16(3.21)	36.22(6.87)	35.89(5.80)	36.37(5.65)	35.59(5.32)
Hip	**M**	-4.15(4.87)	-5.21(3.59)	-4.52(5.21)	-5.61(5.01)	-4.06(5.23)	-5.32(3.93)	-4.81(4.89)	-5.22(5.22)
Y[a]	**F**	-3.54(3.51)	-1.96(4.76)	-1.25(3.49)	-1.66(4.27)	-2.07(3.36)	-1.23(4.61)	-1.03(4.70)	-1.59(4.35)
Hip	**M**	6.73(6.83)	10.20(7.13)	5.38(6.95)	9.03(7.34)	6.16(9.08)	9.46(6.93)	3.62(8.93)	7.47(6.71)
Z[b]	**F**	1.85(9.02)	6.62(11.28)	5.44(8.95)	4.79(6.48)	2.83(6.86)	6.49(6.70)	3.29(8.02)	4.46(5.71)
Knee	**M**	-16.31(9.16)	-17.64(9.39)	-14.46(7.74)	-16.74(8.23)	-16.13(10.56)	-18.08(9.67)	-16.32(9.26)	-17.81(9.11)
X[d]	**F**	-17.93(6.20)	-18.72(4.41)	-18.33(7.80)	-16.06(6.79)	-20.15(3.96)	-18.14(5.46)	-19.36(6.82)	-17.01(6.23)
Knee	**M**	-0.80(2.76)	-0.14(1.05)	0.30(2.31)	-0.43(2.12)	-1.06(2.91)	-0.08(1.54)	-0.03(2.31)	0.32(2.35)
Y[e]	**F**	-0.80(1.97)	-0.27(2.90)	0.09(1.68)	-0.06(2.27)	0.02(2.43)	-0.47(1.82)	-1.35(1.98)	-0.87(2.20)
Knee	**M**	-2.97(2.47)	-3.79(3.47)	-2.72(3.30)	-3.10(3.34)	-3.00(3.38)	-2.44(4.11)	-2.39(3.10)	-3.24(3.55)
Z	**F**	-1.21(3.40)	-2.64(5.46)	-2.06(3.25)	-3.22(3.60)	-2.66(2.99)	-2.46(4.16)	-0.97(3.54)	-0.81(3.58)
Ank	**M**	-5.12(15.97)	0.88(10.15)	-5.84(16.47)	-1.21(14.35)	-4.13(15.27)	0.07(11.38)	-3.48(13.46)	-0.04(11.03)
X	**F**	-0.13(14.32)	-0.91(8.11)	-1.66(12.57)	-2.85(10.15)	-2.97(8.37)	-5.6(10.57)	-4.06(8.49)	-3.92(10.71)
Ank	**M**	3.83(6.75)	3.23(7.72)	4.10(5.60)	3.74(7.63)	2.64(5.84)	4.01(6.62)	4.38(6.48)	4.25(6.04)
Y	**F**	0.46(5.83)	6.36(5.97)	3.20(7.16)	5.83(5.99)	2.61(7.83)	6.25(7.58)	2.79(7.24)	7.02(7.09)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.

Table C.2 Mean (SD) Peak of Stance (0% - 100%) hip, knee and ankle joint rotations (in degrees).

		20 kg		25 kg		30 kg		35 kg	
		Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
Hip X	**M**	36.43(1.78)	35.75(1.39)	36.86(1.43)	38.11(1.21)	35.71(1.67)	36.73(1.36)	36.96(1.36)	38.89(1.54)
Hip X	**F**	36.42(2.21)	36.89(1.72)	35.43(1.77)	37.31(1.51)	37.00(2.08)	36.56(1.70)	37.27(2.03)	36.34(1.91)
Hip Y[a]	**M**	0.17(0.96)	-1.63(1.13)	0.04(1.32)	-0.90(1.25)	0.12(1.05)	-1.56(0.88)	-0.30(1.13)	-1.06(1.05)
Hip Y[a]	**F**	1.56(1.91)	2.45(4.40)	3.30(1.64)	3.70(1.55)	2.17(1.31)	3.75(1.09)	2.26(1.41)	3.60(1.31)
Hip Z[b]	**M**	10.66(1.54)	14.82(2.05)	9.78(1.64)	14.50(1.60)	11.49(1.76)	15.31(1.51)	9.66(1.92)	13.29(1.50)
Hip Z[b]	**F**	7.95(1.91)	10.56(2.55)	10.51(2.03)	9.63(1.99)	7.41(2.19)	11.38(1.88)	8.50(2.39)	10.25(1.86)
Knee X	**M**	-50.59(1.87)	-50.24(1.78)	-50.04(1.41)	-50.83(1.52)	-50.48(1.60)	-51.05(1.54)	-50.56(1.39)	-49.75(1.40)
Knee X	**F**	-53.30(2.33)	-53.34(2.22)	-50.99(1.76)	-51.58(1.89)	-50.54(1.99)	-49.55(1.91)	-49.51(1.72)	-49.66(1.74)
Knee Y	**M**	-3.55(0.70)	-2.26(0.60)	-2.49(0.60)	-2.63(0.63)	-3.28(0.68)	-2.60(0.636)	-2.61(0.59)	-1.93(0.84)
Knee Y	**F**	-3.03(0.86)	-1.98(0.75)	-1.52(0.75)	-2.65(0.78)	-2.41(0.85)	-2.47(0.79)	-3.83(0.73)	-3.02(0.67)
Knee Z[f]	**M**	-5.77(0.88)	-6.13(0.83)	-5.83(0.68)	-6.47(0.78)	-6.49(0.93)	-5.34(0.754)	-6.22(0.86)	-6.10(0.79)
Knee Z[f]	**F**	-7.14(1.09)	-8.93(1.04)	-7.83(0.94)	-8.66(0.98)	-7.73(1.16)	-7.49(0.94)	-6.92(1.07)	-6.65(0.98)
Ank X	**M**	23.03(1.53)	22.95(1.440	21.01(1.90)	23.19(1.340	21.31(1.56)	22.32(1.32)	21.86(1.71)	22.09(1.26)
Ank X	**F**	26.09(1.90)	24.27(1.79)	25.92(2.36)	23.90(1.66)	25.92(1.94)	21.59(1.65)	24.42(2.12)	23.93(1.57)
Ank Y	**M**	2.27(1.83)	1.94(1.56)	2.81(1.27)	1.53(1.52)	1.14(1.49)	2.42(1.46)	2.53(1.38)	2.46(1.16)
Ank Y	**F**	-0.36(2.31)	5.03(1.98)	0.21(1.60)	3.05(1.93)	-0.76(1.88)	3.15(1.84)	0.61(1.75)	4.47(1.46)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.

Table C.3 **Mean (SD) Peak of Stance (0% - 100%) hip, knee and ankle joint moments (in Nm/kg*m).**

		20 kg		25 kg		30 kg		35 kg	
		Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
Hip	**M**	-1.45(0.42)	-1.47(0.30)	-1.63(0.58)	-1.66(0.48)	-1.63(0.52)	-1.66(0.41)	-1.57(0.57)	-1.65(0.41)
X[a,c]	**F**	-1.26(0.27)	-1.23(0.20)	-1.26(0.15)	-1.28(0.22)	-1.30(0.28)	-1.37(0.42)	-1.41(0.30)	-1.48(0.52)
Hip	**M**	-0.80(0.40)	-0.53(0.26)	-0.84(0.40)	-0.58(0.3)	-0.78(0.46)	-0.57(0.32)	-0.79(0.42)	-0.60(0.31)
Y[a,b]	**F**	-1.20(0.34)	-1.18(0.37)	-1.21(0.25)	-1.15(0.35)	-1.23(0.32)	-1.15(0.26)	-1.16(0.41)	-1.10(0.37)
Hip	**M**	-0.39(0.15)	-0.69(0.24)	-0.45(0.14)	-0.67(0.32)	-0.46(0.16)	-0.75(0.23)	-0.45(0.18)	-0.73(0.29)
Z[b,d]	**F**	-0.64(0.35)	-0.61(0.36)	-0.57(0.18)	-0.59(0.31)	-0.62(0.31)	-0.60(0.30)	-0.63(0.23)	-0.62(0.29)
Knee	**M**	2.05 (0.35)	2.00 (0.32)	2.23 (0.45)	2.22 (0.30)	2.20 (0.44)	2.13 (0.39)	2.25 (0.31)	2.12 (0.38)
X[e]	**F**	2.01 (0.14)	2.05 (0.18)	1.91 (0.29)	2.03 (0.28)	1.80 (0.19)	1.96 (0.25)	1.96 (0.24)	2.06 (0.41)
Knee	**M**	0.36 (0.26)	0.34 (0.19)	0.30 (0.20)	0.45 (0.27)	0.38 (0.28)	0.39 (0.22)	0.41 (0.25)	0.43 (0.24)
Y[f]	**F**	0.27 (0.21)	0.22 (0.11)	0.23 (0.14)	0.26 (0.09)	0.24 (0.20)	0.28 (0.12)	0.25 (0.12)	0.37 (0.12)
Knee	**M**	0.28 (0.17)	0.35 (0.16)	0.30 (0.15)	0.39 (0.27)	0.27 (0.18)	0.31 (0.18)	0.26 (0.15)	0.36 (0.19)
Z[a,b]	**F**	0.38 (0.15)	0.62 (0.22)	0.37 (0.07)	0.47 (0.23)	0.35 (0.10)	0.48 (0.21)	0.34 (0.13)	0.49 (0.27)
Ank	**M**	-0.31(0.10)	-0.41(0.20)	-0.29(0.10)	-0.38(0.27)	-0.31(0.10)	-0.47(0.20)	-0.33(0.11)	-0.42(0.22)
X	**F**	-0.41(0.19)	-0.34(0.26)	-0.35(0.12)	-0.36(0.27)	-0.39(0.19)	-0.35(0.26)	-0.40(0.16)	-0.33(0.22)
Ank	**M**	0.04(0.07)	0.03(0.04)	0.05(0.06)	0.05(0.06)	0.04(0.06)	0.05(0.07)	0.07(0.13)	0.04(0.06)
Y	**F**	0.06(0.08)	0.05(0.09)	0.09(0.10)	0.05(0.07)	0.09(0.10)	0.06(0.09)	0.09(0.08)	0.08(0.09)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.

Table C.4 Mean (SD) Coefficient of Variation for Initial Contact hip, knee and ankle joint rotations (in percent).

		20 kg		25 kg		30 kg		35 kg	
		Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
Hip	**M**	6.9 (4.9)	7.2 (5.0)	7.5 (6.2)	6.05 (4.6)	8.4 (5.7)	5.19 (3.0)	9.1 (5.2)	6.8 (3.1)
X[b]	**F**	7.5 (7.9)	6.2 (3.1)	8.5 (4.0)	6.16 (3.2)	6.0 (4.6)	4.88 (2.8)	7.2 (5.7)	6.5 (4.0)
Hip	**M**	336.8 (887.7)	568.8 (2086.3)	55.3 (60.9)	139.9 (232.3)	77.4 (103.6)	183.9 (458.1)	84.2 (96.6)	195.1 (283.0)
Y	**F**	75.0 (49.9)	117.9 (125.3)	90.7 (77.9)	112.1 (180.0)	794.6 (2258.9)	134.7 (123.8)	165.0 (382.9)	93.7 (102.8)
Hip	**M**	56.4 (59.8)	170.0 (349.6)	89.6 (103.3)	86.9 (124.1)	119.0 (246.4)	43.3 (47.2)	395.4 (1383.4)	92.2 (158.5)
Z	**F**	96.3 (119.0)	264.1 (506.8)	44.4 (44.3)	96.87 (84.1)	168.49 (387.1)	123.9 (244.6)	35.0 (25.1)	229.0 (293.2)
Knee	**M**	30.4 (32.5)	26.36 (33.6)	40.2 (45.0)	68.4 (147.9)	28.6 (27.4)	21.2 (11.6)	34.8 (44.0)	38.3 (67.5)
X	**F**	17.9 (11.7)	22.3 (18.8)	15.6 (16.6)	14.3 (8.5)	14.1 (13.0)	21.8 (12.5)	9.7 (10.6)	19.0 (8.4)
Knee	**M**	94.7 (197.5)	166.1 (216.2)	98.8 (182.3)	280.6 (735.6)	82.9 (124.6)	156.3 (197.5)	56.5 (72.1)	224.0 (462.0)
Y	**F**	92.6 (108.8)	35.0 (16.1)	165.1 (296.9)	50.4 (43.1)	89.7 (94.4)	111.3 (118.9)	67.2 (82.4)	88.3 (111.4)
Knee	**M**	86.8 (69.2)	213.6 (708.0)	65.3 (70.9)	89.4 (62.0)	121.7 (170.7)	61.6 (125.7)	74.3 (86.1)	58.7 (80.8)
Z[b]	**F**	113.1 (145.7)	362.0 (877.1)	73.78 (9.4)	454.0 (1211.4)	111.87 (107.3)	271.56 (396.0)	173.9 (239.5)	146.4 (125.7)
Ank X	**M**	31.6 (27.5)	118.7 (157.3)	23.6 (15.8)	119.7 (278.6)	88.1 (117.5)	64.2 (79.5)	44.3 (50.9)	714.9 (2601.75)
	F	252.8 (689.7)	255.5 (377.2)	149.2 (250.1)	185.1 (217.8)	78.2 (75.2)	93.9 (143.5)	77.5 (69.7)	198.16 (411.2)
Ank	**M**	336.8 (887.6)	13.4 (15.0)	107.5 (141.9)	12.8 (6.8)	77.8 (108.0)	13.5 (12.61)	114.3 (233.7)	18.5 (23.1)
Y	**F**	75.0 (49.9)	12.2 (8.8)	65.2 (57.5)	65.7 (135.7)	216.3 (354.8)	12.4 (8.8)	72.3 (73.6)	15.8 (8.6)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.

Table C.5 Mean (SD) Coefficient of Variation for Peak of Stance (0-100%) hip, knee and ankle joint rotations (in percent).

		20 kg		25 kg		30 kg		35 kg	
		Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
Hip	**M**	7.6 (5.7)	8.4 (7.6)	7.7 (7.0)	6.6 (5.1)	8.6 (5.6)	5.3 (3.0)	8.9 (5.2)	7.1 (3.3)
X[b]	**F**	6.2 (3.6)	5.0 (2.5)	8.5 (3.1)	5.2 (2.5)	6.4 (4.4)	4.0 (3.0)	505 (3.8)	5.4 (3.4)
Hip	**M**	188.0 (255.7)	156.1 (234.3)	282.2 (440.0)	119.7 (159.0)	165.6 (223.5)	124.0 (111.4)	223.7 (392.9)	135.7 (180.7)
Y	**F**	124.8 (208.8)	233.5 (579.4)	111.4 (184.8)	52.4 (36.8)	132.6 (148.7)	110.6 (194.8)	125.4 (276.6)	373.4 (1084.8)
Hip	**M**	39.1 (44.0)	25.6 (26.1)	212.8 (745.7)	41.9 (107.2)	78.6 (164.5)	44.2 (123.2)	69.7 (91.1)	25.1 (25.0)
Z	**F**	79.8 (122.4)	52.1 (88.9)	61.9 (109.9)	179.3 (478.3)	74.2 (86.3)	34.0 (27.9)	59.7 (96.0)	43.9 (31.0)
Knee	**M**	5.8 (2.2)	7.2 (4.1)	6.0 (3.8)	6.1 (3.1)	7.4 (3.2)	6.6 (3.5)	6.8 (3.6)	7.3 (4.8)
X	**F**	5.5 (4.1)	6.9 (4.3)	6.1 (2.7)	4.1 (4.6)	6.3 (5.4)	4.6 (2.7)	4.0 (2.0)	6.6 (3.7)
Knee	**M**	63.6 (188.9)	35.1 (32.0)	21.7 (18.2)	30.4 (27.0)	293.2 (983.3)	23.6 (22.6)	66.2 (112.2)	27.9 (52.8)
Y	**F**	26.0 (27.1)	56.8 (96.3)	22.1 (27.7)	28.0 (18.5)	79.9 (190.5)	1183.4 (3419.7)	106.1 (200.2)	21.8 (19.5)
Knee	**M**	30.5 (31.4)	31.0 (17.2)	34.3 (28.1)	30.1 (21.5)	35.1 (31.9)	26.6 (31.4)	70.2 (156.6)	95.5 (227.3)
Z	**F**	35.1 (24.8)	51.4 (101.9)	27.0 (20.2)	13.3 (3.8)	26.14 (18.7)	23.5 (13.2)	21.1 (20.6)	13.5 (8.25)
Ank	**M**	11.0 (7.4)	10.3 (9.2)	10.1 (9.5)	11.8 (5.5)	13.3 (10.2)	11.4 (9.4)	15.5 (14.4)	13.1 (8.7)
X	**F**	10.1 (9.5)	9.7 (5.3)	7.1 (3.3)	13.1 (7.0)	9.6 (4.2)	11.4 (8.7)	7.1 (3.5)	13.4 (7.4)
Ank	**M**	50.8(48.6)	107.2 (169.4)	126.0 (170.2)	138.0 (257.3)	77.1 (92.2)	125.3 (216.3)	73.5 (110.2)	254.6 (700.0)
Y	**F**	295.8 (465.7)	63.1 (58.9)	111.5 (142.2)	106.7 (117.0)	456.7 (169.1)	152.7 (182.4)	77.6 (80.4)	68.1 (77.0)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.

Table C.6 Mean (SD) Coefficient of Variation for Peak of Stance (0-100%) hip, knee and ankle joint moments (in percent).

		20 kg		25 kg		30 kg		35 kg	
		Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
Hip	**M**	16.6 (12.7)	19.3 (14.1)	18.1 (12.4)	14.1 (10.0)	21.1 (14.3)	13.6 (6.8)	12.0 (10.2)	15.4 (15.0)
X	**F**	13.4 (7.8)	14.1 (8.9)	17.3 (6.5)	14.5 (10.3)	10.69 (8.7)	16.7 (9.6)	10.3 (5.0)	18.7 (14.4)
Hip	**M**	23.9 (18.5)	49.1 (66.9)	31.8 (29.1)	43.9 (38.7)	63.5 (108.6)	28.8 (18.0)	46.8 (72.1)	60.3 (125.0)
Y[a]	**F**	9.8 (5.2)	11.7 (8.2)	12.7 (5.5)	12.8 (8.8)	13.3 (8.6)	18.8 (16.2)	15.5 (13.9)	14.4 (11.1)
Hip	**M**	21.1 (17.2)	26.8 (18.0)	19.9 (15.7)	25.7 (16.6)	20.5 (10.3)	24.6 (16.8)	21.4 (12.8)	24.5 (11.9)
Z[b,d]	**F**	11.8 (6.5)	31.2 (24.6)	13.5 (7.7)	32.4 (13.8)	12.8 (11.0)	43.7 (23.3)	16.6 (9.2)	28.4 (13.3)
Knee	**M**	8.1 (4.9)	6.5 (3.1)	7.7 (4.9)	8.0 (5.0)	6.8 (5.2)	8.7 (7.2)	8.1 (5.3)	11.0 (7.5)
X	**F**	7.2 (4.2)	7.9 (3.3)	8.1 (6.2)	9.0 (4.9)	9.6 (7.4)	8.5 (3.2)	6.0 (5.7)	11.2 (14.5)
Knee	**M**	129.4 (262.3)	117.4 (272.3)	39.5 (44.1)	66.7 (56.9)	64.5 (72.3)	160.6 (443.6)	76.0 (114.0)	43.6 (34.1)
Y	**F**	30.2 (22.2)	48.0 (34.6)	25.7 (20.8)	37.7 (22.0)	23.2 (22.1)	140.8 (283.8)	35.1 (22.1)	41.7 (33.1)
Knee	**M**	34.8 (33.0)	46.8 (32.0)	51.4 (71.3)	60.8 (67.2)	49.0 (50.3)	55.7 (44.4)	42.9 (51.7)	54.6 (63.3)
Z	**F**	26.0 (36.3)	27.7 (17.9)	18.7 (14.8)	51.7 (40.8)	16.6 (18.5)	17.8 (12.5)	17.7 (44.8)	44.0 (44.8)
Ank	**M**	6.6 (4.1)	13.4 (11.4)	7.0 (4.4)	8.2 (5.9)	6.7 (4.7)	9.1 (10.9)	7.7 (5.9)	8.0 (9.2)
X	**F**	10.9 (6.0)	7.2 (4.3)	17.2 (26.7)	9.9 (5.3)	5.6 (3.9)	9.9 (5.6)	7.4 (4.6)	6.3 (4.1)
Ank	**M**	50.8 (48.6)	107.2 (169.4)	128.3 (168.7)	105.7 (170.1)	77.1 (92.2)	125.3 (216.3)	73.5 (110.2)	254.6 (700.0)
Y	**F**	295.8 (465.7)	63.1 (58.9)	93.0 (125.3)	68.9 (60.8)	459.6 (169.1)	152.7 (182.4)	77.6 (80.4)	68.1 (77.0)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.

Table C.7 Mean (SD) Coefficient of Variation for Peak Proximal Anterior Tibial Shear (in percent).

	20 kg		25 kg		30 kg		35 kg	
	Dom	**Non**	**Dom**	**Non**	**Dom**	**Non**	**Dom**	**Non**
M	38.1 (28.5)	144.8 (438.7)	35.3 (31.6)	17.1 (185.5)	37.8 (39.7)	36.4 (36.0)	121.0 (589.7)	28.9 (15.9)
F	21.4 (14.7)	39.7 (36.7)	19.7 (14.2)	16.9 (11.5)	18.7 (21.7)	21.9 (20.8)	14.4 (6.6)	28.5 (39.7)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant sex by load interaction.
[f] Denotes a significant limb by load interaction.