EFFECTS OF PROLONGED LOAD CARRIAGE ON KNEE ADDUCTION BIOMECHANICS

by

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The following individuals read and discussed the thesis submitted by student Micah Daniel Drew, and they evaluated their 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: Incidence of knee osteoarthritis (OA) in service members is twice that of the general population. Yet, it is currently unknown how body borne load and duration of walking with body borne load impact knee adduction, biomechanics linked to progression and severity of OA. **Purpose:** This study sought to examine magnitude and variability of knee adduction joint angle and moment throughout a prolonged walking task with body borne load. **Methods:** Eighteen participants had knee biomechanics quantified every five minutes while they walked at 1.3 m/s during a 60-minute over-ground walking task with three body-borne loads (unloaded, 15 kg and 30 kg). **Statistical Analysis:** Thirteen participants with complete data sets were submitted to statistical analysis. Peak of stance (0-100%) knee adduction joint angle and moment, initial contact and range of adduction motion, and coefficient of variation of peak knee adduction angle and moment, and range of adduction motion were submitted to a repeated measures ANOVA to test the main effect and interaction between time (*0, 15, 30, 45 and 60 min*) and load (*0 ,15 and 30 kg*). **Results:** Body borne load significantly increased peak knee adduction moment ($p<0.001$) but not knee peak stance, initial contact or range of adduction angle (all: $p>0.05$); whereas duration of walking task significantly increased peak stance ($p<0.001$) and range of knee adduction motion ($p<0.001$) but not knee adduction moment ($p=0.617$). Neither body borne load nor duration of walking had a significant effect on knee adduction moment, angle or range or motion variability (all: $p>0.05$). **Conclusion:** Prolonged walking with body borne load increased knee adduction

biomechanics related to knee OA pathogenesis. The larger knee adduction moment exhibited with the addition of load and the larger knee adduction angle exhibited towards the end of the prolonged walking task may increase loading of the medial knee joint compartment and increase risk of knee OA.

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LIST OF ABBREVIATIONS

GRF	Ground Reaction Force
BW	Body Weight
IC	Initial contact
PS	Peak Stance
ROM	Range of Motion
KAA	Knee Adduction Angle
KAM	Knee Adduction Moment
MSI	Musculoskeletal Injury
OA	Osteoarthritis
3D	Three-dimensional
CV	Coefficient of Variation

CHAPTER ONE: INTRODUCTION

Musculoskeletal injuries and disease are a substantial problem for occupational athletes such as members of the armed services[1,2]. Musculoskeletal injuries account for more than 55% of all injuries[1,3], and are the leading cause of medical care for service members[2,4,5]. The military spends more than $700 million annually treating musculoskeletal injuries in active-duty service members[3,6]and more than $1.5 billion annually treating the chronic pain and loss of joint function that resulted from musculoskeletal disease in veteran service members[6]. During basic training, upwards of 70% of service members[7,8] suffer a training-related musculoskeletal injury, with most occurring at the knee[1,9–12]. Suffering a training related musculoskeletal injury not only results in loss of duty time, but triples the likelihood of medical discharge and long term disability from subsequent musculoskeletal disease development[9]. In fact, Rivera et al[13] reported that 100% of active-duty service members that suffer a knee musculoskeletal injury develop osteoarthritis (OA), a degenerative musculoskeletal disease at the joint that is the leading cause of medical discharge[5,14–16]. It is reported that incidence of OA in active duty service members and veterans is twice that of the general population[16]. During training activities, service members routinely carry body borne loads, often greater than 30 kg, that reportedly alter lower limb biomechanics increasing risk of musculoskeletal injury[1,16] and might contribute to musculoskeletal disease development[8,10,12], particularly knee OA. However, it is unknown whether walking with body borne load changes specific knee mechanics linked to OA severity and progression.

Knee OA is characterized by degeneration of the joint's articular surfaces, due to wear and tear from repetitive abnormal loading during weight-bearing activities, such as locomotion[17]. During locomotion, the compressive loading of the knee joint is reportedly 2.5 times greater on the medial joint compartment, where knee OA is most prevalent. This is further exacerbated by altered lower limb biomechanics that contribute to knee OA development[18,19]. The external knee adduction moment (KAM) is purportedly a correlate of medial knee joint compartment loading, which is determined by ground reaction force and its lever arm[20–22]. Increases in KAM are related to the severity and progression of knee OA[18,19,21,23,24], where it is reported that each 1% increase in KAM results in 6.5 times faster disease progression[25]. The external KAM also acts to push the knee into valgus and increase peak knee adduction angle (KAA). In fact, patients with radiographically confirmed OA reportedly exhibit significantly greater peak KAA, up to 4 °, during the stance phase of locomotion than healthy controls[26,27].

During locomotion, adding body borne load results in lower limb biomechanics thought to increase musculoskeletal injury and disease risk. Walking with body borne loads increases peak vertical ground reaction forces (GRFs) between 5 and 10%[28–33]. The elevated GRFs are reported to increase lower limb joint and soft tissue loading[34,35] raising the risk of musculoskeletal disease[29,36,37]. To compensate for the elevated GRFs, individuals commonly exhibit significant gait spatiotemporal and joint biomechanical alterations when walking with body borne load. At the knee, individuals increase knee flexion at heel strike and range of flexion motion across stance[28,31,36,38–42]. The increased knee flexion helps the surrounding musculature function as a shock absorber and aids with attenuation of the elevated GRFs placed on the lower limb[41], but leads to larger joint moments[33,39,43,44] which reportedly increases compressive loading of the medial compartment. However, only one study evaluated whether body borne load results in similar changes in KAM, which is more directly related to musculoskeletal disease, but found no change[41] so more research should focus on this potential connection.

Prolonged load carriage may further increase musculoskeletal injury risk. Peak vertical ground reaction forces are reported to increase throughout the duration of walking with body borne load[37,45]. The continual increase in peak GRFs requires ever increasing muscle force to provide adequate joint stabilization[37], accelerating fatigue. The reduced muscle force associated with fatigue limits the lower limb musculature's ability to attenuate the repetitive loading continuously placed on the musculoskeletal system during prolonged load carriage[35,45,46]. This inability and onset of fatigue, present as adaptations in lower limb biomechanics, including further increases in sagittal plane knee range of motion[28,40] and gait variability[33,40]. Natural variability in gait reduces repetitive actions and overuse patterns, and can be used to analyze the motor function of walking for changes indicative of pathogenesis[47–49]. Specifically, increases in stride variability are related to musculoskeletal disease[48] and decreases in knee motion variability is related to reduced performance or the joint's inability to adequately absorb shock[50]. Individuals with moderate to severe knee OA reportedly exhibited greater variability of gait spatiotemporal patterns[20], but decreased variability of knee adduction motion[50]. It is not yet known whether prolonged load carriage has a similar effect on gait variability, specifically of knee motion.

Considering knee OA is a leading cause of medical discharge for service members, it is imperative to identify the specific knee biomechanical adaptations that occur during prolonged load carriage. Although changes in sagittal plane knee biomechanics are well documented when walking with body borne load, little is known regarding the changes in frontal plane biomechanics, in particular magnitude and variability of knee adduction motion and loads, that occur with the addition of body borne load. With that in mind this study seeks to fill this critical gap and provide the military with a better understanding for why service members develop this disease.

Specific Aims

Specific Aim 1:

To examine knee joint adduction during over-ground locomotion with three body borne loads (0, 15, and 30 kg). Specifically, this study will quantify magnitude and variability of knee adduction joint angle and moment, and range of knee adduction while participants walk over-ground at 3.0 mph (1.3 m/s) with three different body borne loads (*0, 15, and 30 kg*).

Hypothesis 1.1

During over-ground locomotion, participants will exhibit a significant increase in peak knee adduction joint angle and moment, and range of knee adduction angle with 15 kg and 30 kg compared the 0 kg load condition.

Hypothesis 1.2

During over-ground locomotion, participants will exhibit a significant decrease in the coefficient of variation in the knee adduction angle and knee adduction joint moment with the 15 kg and 30 kg compared to 0 kg load condition.

Significance

Understanding how body borne load increases magnitude and variability of frontal plane knee biomechanics will aid the military in reducing the likelihood of training and combat related musculoskeletal injuries and disease. Determining whether body borne load leads to maladaptive knee biomechanics, implicated in the pathogenesis of OA, will provide a better understanding for why service members develop this disease.

Specific Aim 2:

To examine knee joint adduction throughout the duration of over-ground locomotion with three body borne loads (*0, 15, and 30 kg)*. Specifically, this study will quantify magnitude and variability of knee adduction joint angle and moment, and range of knee adduction motion, while participants walk over-ground 3.0 mph (1.3 m/s) for 60 minutes with three different body borne loads (*0, 15, and 30 kg).*

Hypothesis 2.1

Participants will exhibit a significant increase in peak knee adduction angle and moment, and range of knee adduction angle at minutes 15, 30, 45 and 60 compared to minute 0 of the prolonged load carriage task.

Hypothesis 2.2

Participants will exhibit a significant increase in the coefficient of variation of peak knee adduction angle and moment at minutes 15, 30, 45 and 60 compared to minute 0 of the prolonged load carriage task.

Significance

Determining how the duration of prolonged load carriage increases magnitude and variability of frontal plane knee biomechanics will provide the military the knowledge necessary to decrease risk of musculoskeletal disease development during training and operational activities.

CHAPTER TWO: LITERATUR E REVIEW

The following section aims to detail load carriage, specifically the 1) military load carriage, 2) injuries related to load carriage, 3) osteoarthritis risk and mechanics, and 4) lower limb biomechanics of load carriage.

Military Load Carriage

History

Throughout history, there is documentation of load carriage, or the transportation by foot of external mass supported on an individuals' body, by military personnel heading into combat. In the ancient world, Assyrian spearman and Greek foot soldiers reportedly carried around 30 kg of armor and weaponry into battle, while Roman soldiers were expected to march around 32 km per day in full gear[51]. This is in contrast to the modern era where loads rarely exceeded 15 kg due to the use of logistical aides such as horses, carts or camp followers[52]. Starting around the Civil War, however, soldiers were required to carry more of their own loads[30], and with the decline of auxiliary transport use came a linear increase in load weight. Much of the increase over recent decades is due to technological advancements in body armor and weaponry[3,52]. Where the average body mass of U.S. male soldiers between the Civil War years and the turn of the century has only increased 15 kg, the average load mass carried has quadrupled[52].

In the early 20th century, research began to look at the effectiveness of soldier load carriage. It was found that heavy loads carried close to the trunk, and concentrated on the upper back were the most practically efficient[53]. Truly, the lowest energy cost comes from carrying loads on the head, which takes extended training[30], and pack frame and hip belts should be used as often as possible to reduce the pressure on the shoulders and back. Due to the swell of research, the British Army recommended soldiers carry no more than 30% of their body weight[54], with the United States following suit with the same recommendation a few decades later[30]. These recommendations were presented as military research continued examining the effects of load on soldier movement and mobility, such as reports that marching distance decreases 2 kilometers for every additional 10 pounds added over 40 pounds, a significant limitation for a military force.

However, despite these recommendations, soldier loads have continued to increase. The U.S. Department of the Army still lists the 30% body weight load as ideal, but present day infantry in the Middle East will carry on average between 29 kg as a fighting load to more than 60 kg during an emergency march, often exceeding 75% of total body weight[3]. Regardless of load, the Army lists standard march rates as roughly 3 miles per hour (1.3 m/s) on a road during the day, and soldiers can cover up to 56 kilometers in a day during forced marches. The Department states that their “primary consideration is not how much soldiers can carry, but rather how much they can carry without reduced combat effectiveness,” which could be detrimental if the long-term effects of loaded marching, such as injury risk, are not considered.

Musculoskeletal Injuries

In Military History

Throughout history it has been shown that improper load carriage, can lead to injuries, lost battles, and failed missions. More than 2,500 years ago, Persian records indicate that when Cyrus the Younger marched his force of 10,000 mercenaries across the ancient world, many soldiers suffered injuries from these prolonged marches, including muscle damage, fractures, blisters and torn ligaments[51]. In 1870 during the Franco-Prussian War, a contingent of 30,000 Prussian Guards marched through what is now present-day Germany and lost more than a third of their men due to fatigue from load carriage[51].

Incidence in Military

Presently, musculoskeletal injuries (MSIs) are still one of the biggest risks to members of the armed services[16] and the leading cause of medical care[4]. Injury rates in general have tracked with load increases—up more than 700% in the last 25 years[53]—of which musculoskeletal injuries comprise 55%[1,3]. Disabilities due to injuries in general are shown to have a high economic cost for the armed forces[6]. Compensation for injuries and costs of medical care increase every year. It is estimated that costs associated with treating injuries approaches $750 million each year[3], while direct compensation for disabilities, led by lower back and knee conditions, has been as high as $1.5 billion per year[6]. During a two year study among Air Force recruits, 12.5% of recruits sustained MSIs costing $44 million to treat, and those injured were three times as likely to be discharged, or three times as likely to graduate late[9]. This is a conservative figure, as other studies have shown that incidence of MSIs has been estimated at between 19% to 40% for men and between 40% and 70% for women during basic training[7,8].

The lower extremity is the most common site for overuse injuries among trainees and active military members, with nearly 80% occurring there[9]. Within that, a plurality of lower limb injuries occur at the knee, making up of almost half of all non-combat MSIs[1,8,10,12,53]. Marching, patrolling, and combat training have been reported as the most common activities performed at the time of injury[12], and carrying body borne loads greater than 30 kg has reportedly increased incidence of MSIs by more than 100%. Suffering a training related MSI not only results in in loss of duty time but significantly increases the likelihood of long term disability from subsequent musculoskeletal disease development[9]. In fact, Rivera et al[13] reported that 100% of active-duty service members that suffer a knee musculoskeletal injury develop osteoarthritis (OA), a degenerative musculoskeletal disease that is the leading cause of medical discharge from the armed services[5,16].

Musculoskeletal Disease

Osteoarthritis

Osteoarthritis is a progressive degenerative disease that affects the soft tissues, articular cartilage, and bone around joints. The National Institute of Health reports that an estimated 27 million American adults are affected by this chronic disease[16] making it the leading cause of disability for adults[55], and it is often associated with elderly and overweight populations. However, recent studies show that OA, specifically knee OA, is becoming prevalent among young people who engage in physically demanding tasks[56,57], such as members of the armed services[58,59]. In fact active duty service members, veterans and other occupational athletes such as wildland firefighter are shown to have incident rates of OA twice as high as the general population[16]. Research indicates that physically demanding occupational tasks, such as repetitive lifting and carrying heavy loads, squatting and kneeling, can increase the likelihood of systematic knee OA by 30% - 60%, and bring on symptoms at earlier ages[16,56]. In fact, among military populations, incidence of OA in individuals under the age of 40 is significantly greater than that of the general population[59,60].

Knee OA Biomechanics

In all populations, the knee joint is most commonly affected with OA[21]. Knee OA represents the failure of the tissues to repair joint damage[14] due to wear and tear from repetitive abnormal loading during weight bearing activities such as locomotion[17]. During locomotion, the compressive loading of the knee joint is reportedly 2.5 times greater on the medial compartment, where knee OA is most prevalent. It is thought biomechanical changes to the knee joint that lead to increased uneven knee joint loading contribute to the development and progression of knee OA[21]. When ground reaction forces that occur during locomotion increase, there is a direct increase in medial compartment loading[34,35,61]. A common correlate of the medial compartment loading is the external knee adduction moment (KAM), a product of ground reaction force and the moment arm at the knee joint[62,63]. Increases in KAM are shown to relate to the severity and progression of medial knee OA[18,23,24]. In fact, each 1% increase in KAM reportedly results in a 6.5 times faster disease progression[25]. Studies have also shown a relationship between increased KAM and any incidence of knee OA, reporting that patients affected by knee OA exhibit significantly higher peak KAM than unaffected individuals[19,64]. The external KAM also acts to push the knee into valgus and alter angular knee motion[19]. Patients with OA exhibit decreased sagittal plane knee range of motion shown to decrease, while knee adduction range of motion significantly increasing. Patients with knee OA reportedly have an average knee adduction angle that is 5 degrees higher throughout the stance phase of locomotion than healthy controls[23,26,27].

Effects of Load Carriage

Physiological

Carrying excess weight, through a backpack, rucksack or other form, alters biomechanical and physiological parameters during locomotion. Physiologically, walking with load increases heart rate, ventilation, oxygen uptake, metabolic cost, relative work intensity[39] and cost of transport[65,66]. Metabolic costs are found to be 30% to 45% higher when walking with a backpack, with changes starting at loads of just 15% of body weight. Even when pace is adjusted to equalize other variables, walking while loaded is described as being more difficult than while unloaded. In general, energy expenditure increases in proportion to load, however that is directly tied to walking speed and position of the load. Carrying load close to the center of mass, and higher on the back decreases the metabolic cost, and studies have shown that trained individuals can carry up to 60% of their body weight with no change in metabolic cost if the load is placed on the head[30,66].

Spatiotemporal Changes

During locomotion, adding body borne load results in lower limb biomechanics that are thought to increase musculoskeletal injury and disease risk. Spatiotemporal changes to gait are shown to change with loads as low as 8 kg[38]. Walking is characterized by two periods during the gait cycle—the double and single support phases—when both feet are on the ground (double support) or single support where one leg is swinging through the air (swing phase). One of the main gait alterations from increased load is an increased time spent in the double support phase, a decreased length of the stride and the coinciding increase in stride frequency[31,36,38,44]. However, it is usually found that stride length and stride frequency are most altered during a fixed pace, while self-selected pacing during load carriage does not seem to have the same effect[38]. The increased double support time allows individuals to attenuate the higher levels of ground reaction force experienced from the addition of body borne load[28,29,42,67].

Ground reaction forces

Ground reaction forces (GRFs) can offer insight on gait and impact forces acting on the lower extremities, and changes in GRFs are a common measure examined in load carriage literature. It is commonly reported that both vertical and anteroposterior GRFs during gait increase when a load is carried during locomotion, specifically that vertical GRFs increase proportionally with increased loads[28,29,31,66,67]. Across the walking gait cycle, the vertical impact peak, minimum vertical GRF and vertical thrust peak have been repeatedly shown to increase between 5-10%, with some studies showing increases proportional to additional of load[28,29,44]. In the anteroposterior direction, maximum breaking and propulsive forces also increase proportionally with load[29,37]. The elevated GRFs are reported to increase lower limb joint and soft tissue loading[34,35], and increase mediolateral joint stability, raising the risk of musculoskeletal disease[29,36,37].

Trunk and Hip Kinematics

To compensate for the elevated GRFs, individuals commonly exhibit changes to knee, hip and trunk biomechanics during prolonged load carriage. A primary response to load carriage in a pack is an anterior lean of the trunk and head[53]. Trunk lean can occur with as little as 6 kg of weight addition and lead to increased muscle activity in the pelvis and low back to increase postural stability and offset the migration of the center of mass. As the load carried increases, hip range of motion increases during walking, however the effects are not consistent throughout the literature. Hip range of motion is shown to change with an additional load between 0 kg and 7.5 kg, or 15 % increase in bodyweight, however this is not always the case as Birrell[41] and Attwells[38] have reported no range of motion changes between 0 kg and 15 kg and 0 kg and 32 kg respectively. Peak hip flexion has been shown to increase linearly with any load, with significant increases with as little as 7.5 kg[38,40], and up to 40 kg with no additional increases beyond that. Hip angle values at initial contact are also shown to increase with a 15 % body weight addition[67].

Knee Kinematics and Kinetics

Similarly, knee rotational changes in the sagittal plane are not consistent throughout load carriage literature. Knee flexion range of motion increases with loads above 15 kg, but Qu et al and Atwells et al, and Birrell did not find any changes with a 15 kg compared to 0 kg load[38,40]. There have also been instances where knee ROM decreases across the load spectrum[41,68]. What is more consistently reported is an increased sagittal knee angle at initial contact, which is reported with loads starting at 15 % of body weight and above[28,67]. Birrell[41] notes that increased knee flexion upon contact is a function of shock absorption, to counter the higher impact GRFs found with increased load carriage. Few studies have examined frontal plane knee rotations during locomotion, with Birrell[41] finding no changes during walking with load, and Brown reporting that individuals increased knee adduction range of motion and peak values while running with 30% of body weight[69]. Kinetic changes at the knee have been observed with more consistency during load carriage tasks. Knee flexion joint moments show significant increases between the addition of 0% and 15% body weight and 0% and 30% body weight[33,39,67,68,70] during locomotion with a body borne load. Again, the majority of studies only examined flexion-extension moments, neglecting to look at alterations in knee adduction moment, with just Brown reporting increases to KAM during running with body borne load[69].

Fatigue

Prolonged physical activity, specifically prolonged load carriage, is shown to increase fatigue[40,45,71]. With the increased peak ground reaction forces that occur during locomotion with body borne load, muscles are required to produce increased force to provide adequate joint stabilization[37], which accelerates the onset of fatigue. Fatigue reduces muscle force, limiting the lower limb's ability to attenuate the repetitive loading on the musculoskeletal system during prolonged load carriage[35,37,45,46], resulting in significant changes to knee biomechanics, including further increases in sagittal plane knee range of motion[28,40] and gait variability[33,40]. Variability in gait is natural, as it reduces repetitive actions and overuse patterns, and changes in gait variability is often used to analyze the motor function of walking for changes indicative of pathogenesis[47–49]. Specifically, increases in stride variability is related to musculoskeletal disease[48] and decreases in knee motion variability is related to reduced performance or the joint's inability to adequately absorb shock[50]. Individuals with moderate to severe knee OA reportedly exhibited greater variability of gait spatiotemporal patterns[20], but decreased variability of knee adduction motion[50].

Summary

Historically, soldiers have been required to carry heavy loads while marching. In recent years there has been a dramatic rise in the weight soldiers are required to carry, coupled with an increase in injuries among military populations. Research indicates that lower extremity biomechanics are altered while walking with large body borne loads. In addition, altered mechanics due to fatigue and fixed-cadence marching may pose additional injury risk, especially when carrying body borne loads. However, much of the research into biomechanical adaptations to load carriage focuses exclusively on the sagittal plane, while little to no research exists directly examining frontal plane gait changes due to load and duration. As changes in frontal plane knee biomechanics such as peak joint moments and peak joint angle have been previously found to occur during load carriage, it is possible a relationship exists between similar military activities and the high prevalence of OA discharges from the military. This work seeks to further examine this relationship and determine the effects of body borne loads (15 kg and 30 kg) and prolonged walking on knee adduction biomechanics.

CHAPTER THREE: MANUSCRIPT

Introduction

Musculoskeletal injuries are a substantial problem for the military and leading cause of medical care for service members[1,2,4,5]. Annually, the military spends more than $700 million dollars treating the nearly 70% of soldiers who suffer a musculoskeletal injury during service[3,6–9]. A majority of these musculoskeletal injuries occur at the knee during basic and advanced training where soldiers are required to walk for long periods of time with heavy body borne loads (i.e. greater than 30 kg[3])[1,9–12]. Considering, Rivera et al reported that 100% of active-duty service members that suffer a knee musculoskeletal injury develop joint osteoarthritis (OA)[13], a degenerative musculoskeletal disease that is the leading cause of medical discharge and long term disability for service members[5,14–16], it is imperative to understand how walking with body borne load leads to knee musculoskeletal injury and OA development.

Knee OA is characterized by degeneration (i.e., wear and tear) of the joint's articular surfaces from the repetitive application of abnormal load during weight-bearing activities, such as locomotion[17]. Knee OA is most prevalent in the medial joint compartment, and results from compressive joint loads during locomotion that are reportedly 2.5 times greater on the medial compartment than the lateral compartment[61,72]. During locomotion, the external knee adduction moment (KAM), is a correlate of medial knee joint compartment loading that is related to the severity and progression of knee OA[18–24]. Each 1% increase in KAM, in fact, results in a six-fold increase in the rate of knee OA progression[25]. Moreover, KAM pushes the knee into varus and increases peak knee adduction angle (KAA). Peak KAA exhibited during locomotion is purported to be up to 4 degrees greater in patients with radiographically confirmed OA than healthy controls[26,27]. But, it is currently unclear whether walking with body borne load, as commonly done during military training, produces significant increases in knee adduction biomechanics (i.e., KAA and KAM) related to knee OA development.

During locomotion, the addition of heavy, military relevant body borne load results in lower limb biomechanics thought to increase musculoskeletal injury risk and may accelerate progression of musculoskeletal disease[1,16]. Walking with body borne load increases peak vertical ground reaction forces (GRFs) up to 10%[28–33]. These elevated GRFs are reported to increase lower limb joint and soft tissue loading[34,35], potentially placing abnormal loads on the medial knee joint compartment[29,36,37]. To compensate for the elevated GRFs, individuals exhibit lower limb, in particular knee, biomechanical adaptations. Specifically, when walking with body borne load, individuals increase magnitude and range of knee flexion[28,31,36,38–42]. The flexed knee helps the lower limb musculature absorb and attenuate the elevated GRFs placed on the musculoskeletal system[41], but may contribute to significant increases in lower limb joint moments evident when walking with load[33,39,43,44]. Individuals, in fact, are reported to increase knee flexion moments between 10% and 36% when walking with body borne loads[44,68,73]. Both walking and running with body borne load are purported to increase KAM[69,73,74], and may act to further push the knee into adduction. Yet, it is unclear whether heavy military body borne loads produce similar increases in knee adduction biomechanics, in particular KAA when walking for extended periods of time as is common during military training[74].

Prolonged walking with body borne load may further increase knee adduction biomechanics related to musculoskeletal disease. Peak vertical ground reaction forces are reported to increase approximately 2% every fifteen minutes of walking with body borne load[45]. The continual increase of GRFs during prolonged walking with body borne load may require ever-increasing muscle forces to provide adequate joint stabilization, accelerating muscular fatigue and associated weakness[37]. Muscle weakness associated with fatigue may limit the lower limb musculature's ability to attenuate repetitive loading[35,45,46], and present as significant adaptations of knee biomechanics. Specifically, individuals are reported to increase knee flexion range of motion[42] approximately 5% following a fatiguing exercise, but similar increases in knee adduction were not reported during short bouts of loaded walking following fatigue[74]. Muscular weakness may also impact variability of lower limb biomechanics[33,40] increasing risk of musculoskeletal injury[75] and disease[47–49]. Decreased variability of spatiotemporal and joint kinematic measures during walking may increase risk of musculoskeletal disease development in general[55], and decreased variability of knee adduction motion[20] may increase severity of knee OA specifically[50]. Yet to date, it is not known whether prolonged load carriage impacts variability of knee adduction biomechanics and risk of musculoskeletal disease.

Considering knee OA is a leading cause of medical discharge for service members, it is imperative to determine whether prolonged walking with heavy military relevant body borne loads, a common training-related task, produces knee biomechanics related to the risk of OA development. Although changes in sagittal plane knee biomechanics while walking with body borne load are well documented, it is unknown if similar changes in knee adduction biomechanics, in particular magnitude and variability of knee adduction joint angle and moment occur during prolonged load carriage. With that in mind, the purpose of this study was to determine the adaptations in knee adduction exhibited during a prolonged walking task with body borne loads (0, 15 and 30 kg) commonly worn during military training. It is hypothesized that the addition of body borne load and duration of walking would produce significant increases in the magnitude of knee adduction joint angle and moment, while the variability of knee adduction would decrease with body borne load but increase with the duration of walking.

Methods

Participants

An *a priori* power analysis of peak knee adduction moment during similar load carriage tasks indicated a minimum of 16 participants were needed to achieve 80% statistical power with an alpha level of 0.05. We recruited eighteen healthy and recreationally active participants (12 male: 23.3 ± 1.8 yrs, 1.8 ± 0.1 m, 77.9 ± 9.5 kg; 6 female: 22.8 ± 1.8 yrs, 1.7 ± 0.1 m, 59.9 ± 3.5 kg). Each potential participant self-reported the ability to safely carry up to 75 pounds while walking, physical activity level using a PAR-Q (Appendix A)[76], and their injury history (Appendix B). Participants were excluded for having: (1) history of surgery in the low back or lower extremities; (2) pain and/or injuries located in the back or lower extremities in the last six months; (3) any known neurological disorders; and/or (4) were currently pregnant. Research approval was obtained from the local Institutional Review board and all participants provided written informed consent prior to testing.

Experimental Design

Each participant performed one orientation and three test sessions. During each test session, participants completed a prolonged load carriage task with a different body borne load (0 kg, 15 kg, and 30 kg) (Figure 3.1). For each body borne load, participants wore spandex shorts and a shirt. For the 15 kg and 30 kg loads, participants also wore a weighted vest (V-MAX, WeightVest.com, Rexburg, ID, USA) that was systematically adjusted to provide the necessary weight for each condition. Prior to testing, each load configuration was weighed and loads within 2 % of the target were accepted. Each test session was separated by a minimum of 24 hours to minimize fatigue effects and reduce chance of injury. To avoid bias and confounding data, a 3 x 3 Latin square approach was used to randomly assign the load configuration order prior to testing (Table 3.1).

Figure 3.1. Weighted vest set up for each body borne load condition (15 and 30 kg)

Table 3.1. The Latin Square Design used for Randomization of the Testing Order for Each Weighted Condition.

	SESSION 1	SESSION 2	SESSION 3
Order 1	0 kg	15 kg	30 kg
Order 2	15 kg	30 kg	0 kg
Order 3	30 kg	0 kg	15 kg

Experimental Protocol

Orientation Session

The orientation session was used to collect participant demographic and strength data, and to familiarize participants with the different load configurations and testing procedures. During the orientation session each participant had their demographic information, including height (m), weight (kg), age (years), and foot dominance via the Waterloo Footedness Questionnaire (WFQ-R) recorded (Appendix C)[77]. Each participant also had trunk and lower limb strength recorded. To record trunk strength, each participant performed a flexor endurance, modified Biering-Sorensen and side bridge test according to McGill et al[78,79]. To record lower limb strength, each participant completed maximal isometric hip and knee flexion and extension, hip abduction, and ankle dorsiflexion and plantarflexion contractions with their dominant limb on an isokinetic dynamometer (HUMAC NORM , CSMI, Stoughton, MA, USA.). For hip flexion and extension, participants stood with the hip flexed at 15 degrees. For hip adduction participants lay on their non-dominant side with the hip abducted 15 degrees[80,81]. For knee flexion and extension, participants were seated with hip flexed at 85 degrees, thigh secured and knee flexed to 60 degrees[82]. For ankle plantar/dorsiflexion, participants lay prone with the ankle neutral (0 degrees of plantar flexion[81]). Participants were asked to perform three repetitions of each isometric contraction with a 40-second rest period between each repetition[83]. The maximum torque production was recorded for each trial. To familiarize themselves with the testing procedures and body borne loads, participants walked at 1.3 m/s through the motion capture area with each load condition. Each participant was required to give verbal confirmation that they could safely carry the body borne loads and perform the study tasks before testing.

Biomechanical Testing

During each test session, participants had three-dimensional (3D) lower limb (hip, knee and ankle) biomechanical data recorded during the prolonged load carriage task. Ground reaction force (GRF) data (2400 Hz) was collected from one in-ground force platform (AMTI OR6 Series, Advanced Mechanical Technology Inc., Watertown, MA), while eight high-speed (240 Hz) optical cameras (MXF20, Vicon Motion Systems LTD, Oxford, UK) recorded lower limb motion data. Vicon Nexus (v2.6, Vicon Motion Systems, LTD, Oxford, UK) recorded and stored biomechanical data for post processing.

The prolonged load carriage task required each participant to walk over-ground at 1.3 m/s for 60 minutes. Specifically, each participant started indoors at minute 0 and completed one lap (both the indoor and outdoor portions) of the 390-meter walking course (Figure 3.2) every five minutes thereafter (minutes 5, 10, 15…60). For the indoor portion, participants walked 1.3 m/s ± 5 % three times through the motion capture volume. During each walk trial two sets of infrared timing gates (TracTronix TF100, TracTronix Wireless Timing Systems, Lenexa, KS), placed four meters apart in the capture volume quantified walking speed. Each trial was marked as either successful or unsuccessful. A trial was successful if the participant walked the required speed and contacted the force platform with only his or her dominant limb. Throughout the walking task, a metronome was set to the participants' predetermined cadence to ensure correct walking speed.

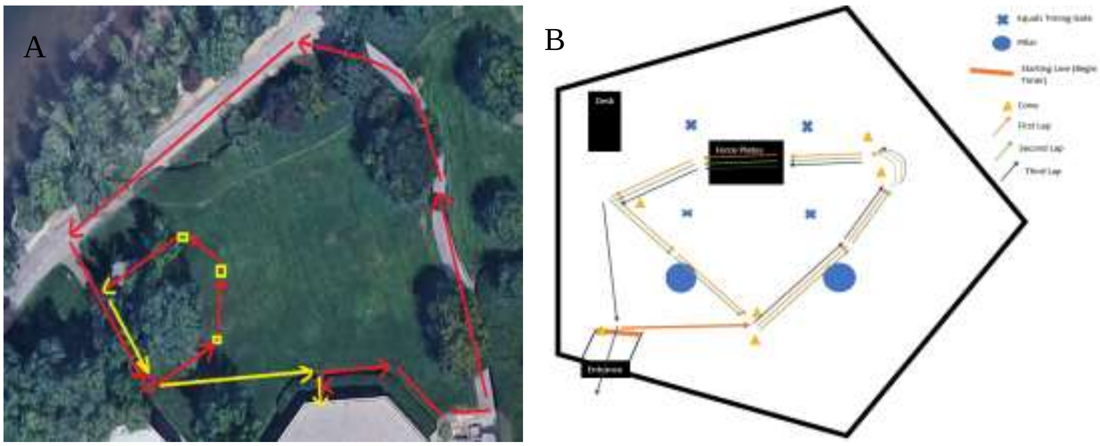

Figure 3.2. The outdoor (A) and indoor (B) loops used during the load carriage walk task

Data Analysis

<u>Biomechanical Analysis</u>

During each trial, lower limb biomechanical data was quantified using the 3D coordinates of 34 retro-reflective and four virtual markers (Table 3.2). Each reflective marker was attached to a specific bony landmark using double-sided tape and secured using elastic tape (Cover-Roll Stretch, BSN Medical, Charlotte, NC, USA). Each virtual marker was created by digitizing a specific bony landmark in the global coordinate system using a Davis Digitizing Pointer (C-Motion Inc., Rockville, MD). After each marker was secure, participants stood in anatomical position for a static recording which was used to create a kinematic model. The kinematic model consisted of eight segments, including trunk, pelvis and bilateral thigh, shank and foot, with 27 degrees of freedom. Each segment had a local coordinate system and three orthogonal axes (x, y and z) assigned in Visual 3D (v6, C-Motion, Inc, Germantown, MD, USA). The trunk was assigned a local coordinate system with three degrees of freedom and joint centers defined at the intersection of the midpoint of the acromion processes and the seventh cervical vertebrae and sternum jugular notch. The pelvis was assigned a local coordinate system with three rotational and three translational degrees of freedom, and a joint center defined halfway between the right and left anterior superior iliac spines. The hip was assigned a local coordinate system with three degrees of freedom and a functional joint center determined according to Rozymalski and Schwarts[84]. The knee and ankle were assigned three degrees of freedom according to Grood and Suntay, and Wu, and joint centers located at the midpoint between the medial and lateral femoral epicondyles and the medial and lateral malleoli, respectively[85,86].

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

	Markers
Trunk	*Acromion process*, **jugular notch, xiphoid process, midpoint between inferior angles of scapulae, C7 vertebrae**
Pelvis	*Anterior-superior iliac spines, posterior-superior iliac spines*, and iliac crests
Thigh	Greater trochanter, distal thigh, *medial* and *lateral femoral epicondyles*
Shank	Tibial tuberosity, lateral fibula, distal tibia, *Medial* and *lateral malleoli*
Foot	Posterior heel, midpoint of first and fifth metatarsal heads, *first and fifth metatarsal heads*

Note: *Italic* indicates calibration markers. **Bold** indicates virtual markers

The synchronous GRF and marker trajectory data for each trial were low pass filtered using a fourth-order Butterworth filter (12 Hz). The filtered marker trajectories were then processed in Visual 3D to calculate knee rotations that were expressed with respect to each participants' static pose using a joint coordinate systems approach[85,87]. Using a standard inverse-dynamics analysis, filtered kinematic and GRF data were processed to obtain 3D forces and moments at each lower limb joint, with segment inertial properties defined according to Dempster et al[88,89]. Knee joint moments were expressed as flexion-extension, abduction-adduction and internal-external rotation and reported as external moments. Joint moments were normalized by body mass (kg) and height (m), and GRFs were normalized to subject body weight (N). All biomechanical data was normalized from 0% to 100% of stance phase and resampled to 1% increments (n = 101). Stance phase was identified as heel strike to toe-off and defined as the moment when GRF first exceeded and fell below 10 N, respectively.

Statistical Analysis

Statistical analysis only included 13 participants (9 male: 23.4 ± 1.5 yrs, 1.8 ± 0.1 m, 77.7 ± 25.5 kg; 4 female: 24.5 ± 4.4 yrs, 1.7 ± 0.1 m, 59.3 ± 4.0 kg) as five participants had insufficient marker data due to technological difficulties or obfuscated markers during testing. Predefined knee biomechanics related to progression and severity of OA were submitted to statistical analysis[19]. Specifically, the kinematic dependent variables included initial contact (IC) and peak of stance (PS, 0%-100%) KAA and range of knee adduction motion (ROM KAA, PS minus IC). The kinetic dependent variable included PS KAM. Each dependent variable was averaged across two successful trials recorded at minutes 0, 15, 30, 45 and 60 of the prolonged walking task to create a participant based mean. Each participant-based mean was submitted to a RM ANOVA to test the main effects of and interaction between load (*0, 15 and 30 kg*) and time (*0, 15, 30, 45 and 60 minutes*). Within subject variability (i.e. Coefficient of Variation (CV)) for

PS KAA, KAM, and ROM KAA was calculated as the standard deviation of the two selected trials divided by the means of those trials ($CV = \sigma/\mu * 100$) and submitted to a similar RM ANOVA[90]. For analyses where sphericity was significant, the Greenhouse-Geisser correction was applied to the degrees of freedom. Significant interactions were submitted to simple effects analysis and a Bonferroni correction was used for pairwise comparisons[91,92]. Alpha was set to *a priori* at $P<0.05$. All statistical analysis was performed using SPSS software (v25 IMB, Armonk, NY, USA).

Results

No significant interactions were observed ($p>0.05$), therefore only main effects are presented below.

Magnitude of Knee Biomechanics

Body borne load had a significant effect on PS KAM ($p<0.001$) (Figure 3.3 and Appendix D). Specifically, PS KAM increased with 30 compared to 15 ($p<0.001$) and 0 kg ($p=0.007$) loads, and with the 15 compared to 0 kg load ($p=0.025$). Body borne load had no effect on IC ($p=0.459$), PS ($p=0.869$) or ROM KAA ($p=0.978$) (Figure 3.4 and Appendix D).

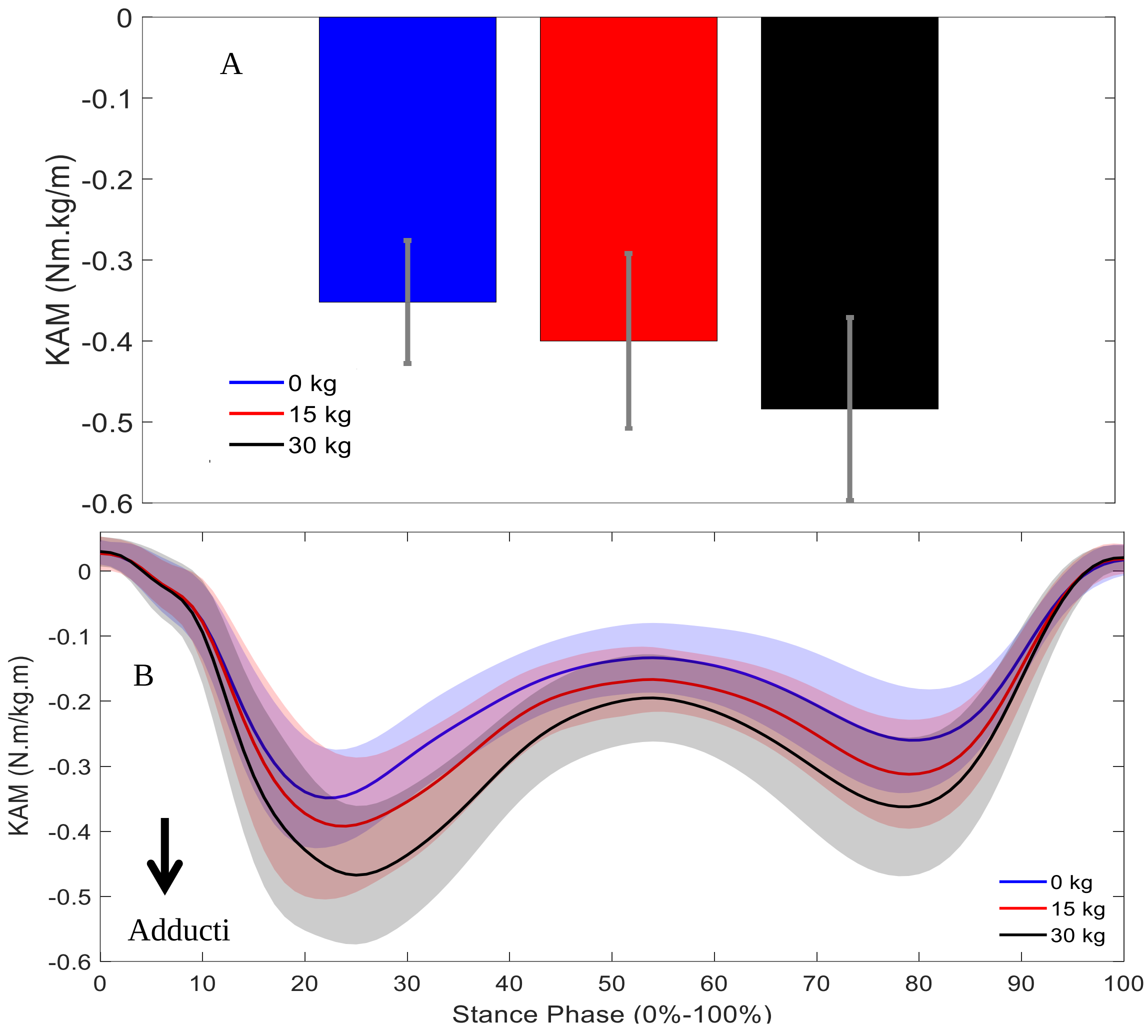

Figure 3.3. Mean peak (A) and stance phase (0% - 100%) (B) knee adduction joint moment during walking task for each body borne load (0, 15 and 30 kg).

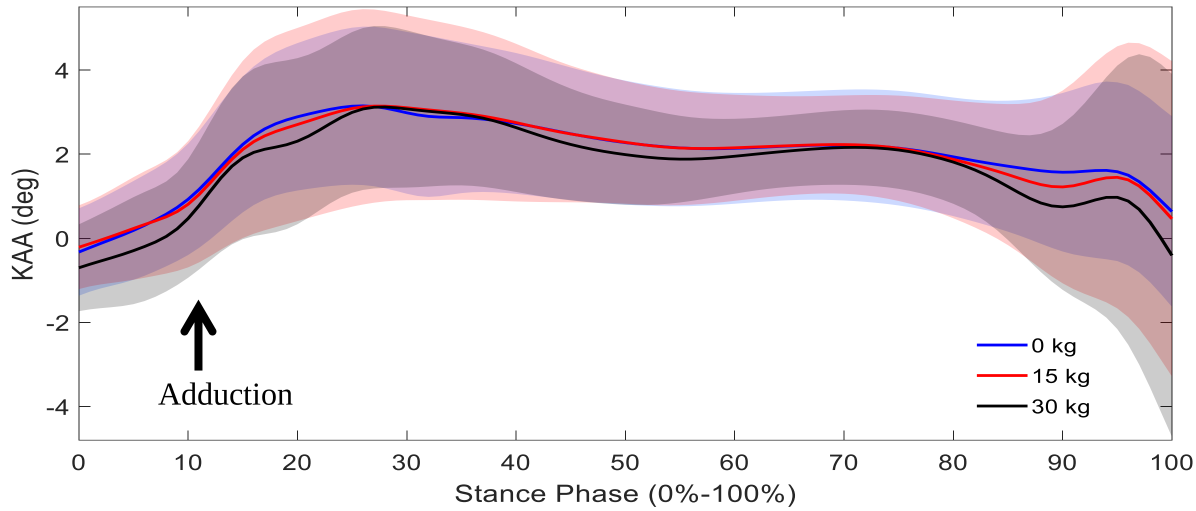

Figure 3.4. Stance phase (0% - 100%) knee adduction joint angle during walking task for each body borne load (0, 15 and 30 kg).

Time had a significant effect on PS ($p<0.001$) and ROM KAA ($p<0.001$) (Figure 3.5 and Appendix D). Specifically, PS KAA was greater at minutes 30 through 60 compared to minute 0 (all: $p<0.005$) and at minute 60 compared to minute 15 ($p=0.030$), while ROM KAA was greater at minute 30 ($p=0.040$) and minute 60 ($p=0.020$) compared to minute 0. No significant difference in PS or ROM KAA was observed between any other times ($p>0.05$). Time had no effect on IC KAA ($p=0.115$) or PS KAM ($p=0.617$) (Figure 3.6 and Appendix D).

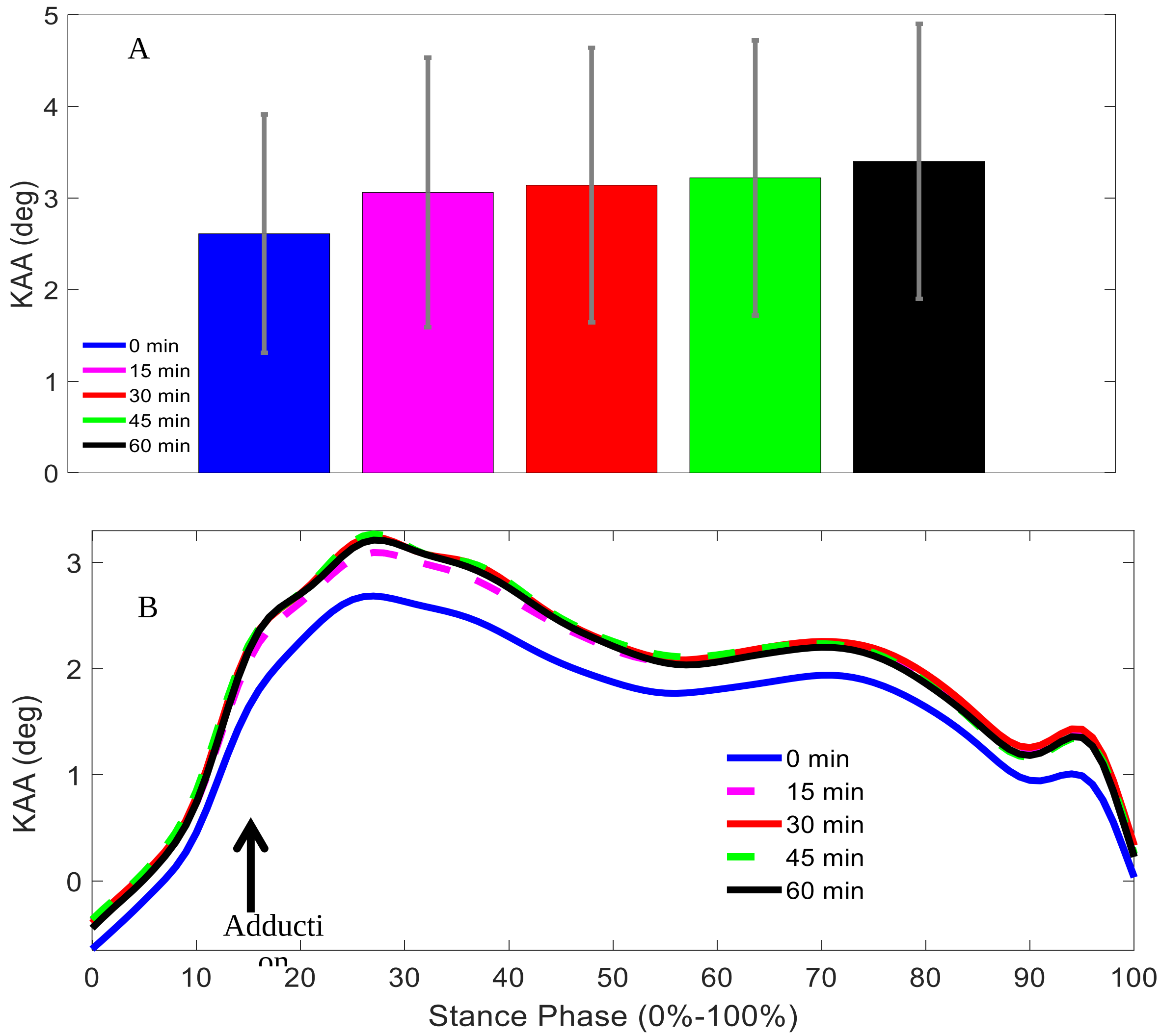

Figure 3.5. **Mean peak (A) and stance phase (0% - 100%) (B) knee adduction joint moment during walking task for each body borne load (0, 15 and 30 kg).**

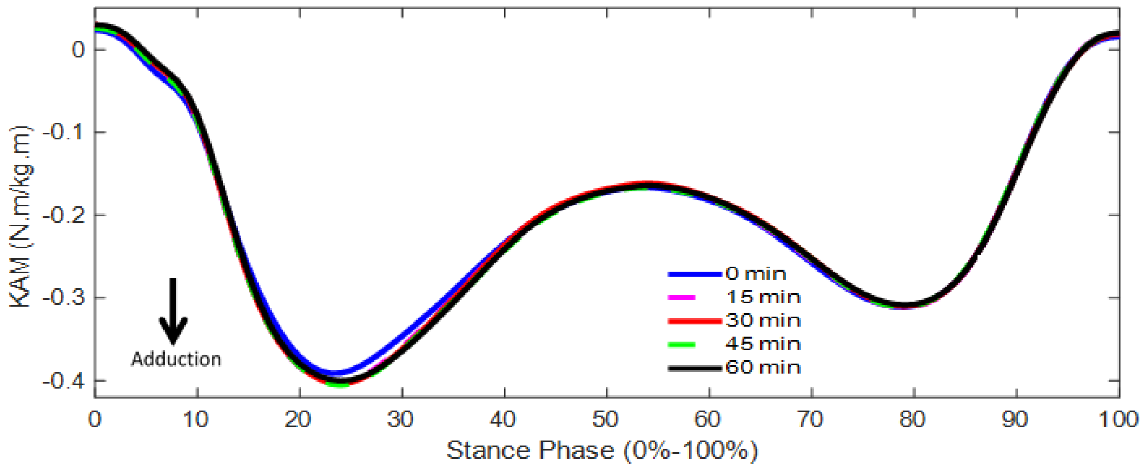

Figure 3.6. Stance phase (0% - 100%) knee adduction joint angle during walking task for each body borne load (0, 15 and 30 kg).

Variability of Knee Biomechanics

Neither body borne load, nor time had a significant effect on CV of KAA (p=0.319; p=0.302), KAM (p=0.645; p=0.485) or ROM (p=0.476; p=0.412) (Table 3.3).

Table 3.3. Mean variability (CV) of peak knee adduction joint angle, moment and range of motion with each body borne load a minutes 0 through 60 of prolonged walking task.

		0 Min	15 Min	30 Min	45 Min	60 Min
KAM CV	**0 kg**	24.58 ± 65.15	10.26 ± 18.26	9.11 ± 12.38	7.65 ± 8.56	13.82 ± 14.04
	15 kg	74.39 ± 206.6	5.61 ± 5.47	17.55 ± 31.84	63.65 ± 200.00	9.95 ± 7.18
	30 kg	9.28 ± 8.53	9.35 ± 8.66	9.28 ± 7.54	9.58 ± 13.43	10.42 ± 11.89
KAA CV	**0 kg**	5.73 ± 4.41	6.28 ± 7.27	6.25 ± 4.81	4.94 ± 3.45	6.02 ± 4.58
	15 kg	10.45 ± 11.80	5.08 ± 3.05	6.83 ± 3.33	4.82 ± 5.84	4.65 ± 3.60
	30 kg	5.02 ± 5.01	7.20 ± 8.39	4.59 ± 2.85	5.75 ± 6.20	5.34 ± 2.86
ROM CV	**0 kg**	12.30 ± 14.59	10.45 ± 17.48	10.45 ± 7.90	9.08 ± 5.53	10.12 ± 10.51
	15 kg	15.10 ± 15.08	7.45 ± 5.84	9.83 ± 9.34	7.56 ± 6.90	8.19 ± 4.66
	30 kg	8.51 ± 6.93	8.10 ± 5.98	9.57 ± 7.00	9.82 ± 8.11	8.11 ± 10.59

Discussion

This study sought to examine whether walking for 60-minutes with heavy, military relevant body borne loads (0, 15 and 30 kg) increased magnitude and variability of knee adduction biomechanics related to musculoskeletal injury and disease. Our hypotheses, however, were only partially supported as magnitude, but not variability, of knee adduction increased with body borne load and walking time respectively.

Walking with body borne load may produce knee biomechanics that increase the likelihood of developing musculoskeletal disease at the joint. In agreement with previous literature, peak KAM increased 0.05 and 0.13 Nm/kgm when walking with the 15 and 30 kg body borne loads, respectively[73]. Increases in the external knee adduction moment reportedly load the medial knee joint compartment, and may accelerate the wear and tear of the joint's articular surfaces that lead to knee OA[64,93]. In fact, patients with radiographically confirmed knee OA exhibit up to 30% higher peak KAM than healthy controls[19,26]. Considering the current participants increased peak KAM approximately 37% with the 30 kg body borne load it is possible that routine military training activities, such as walking with heavy body borne loads, may increase knee biomechanics implicated in OA development. Despite the large increases in KAM with the addition of heavy body borne load, there was not a significant, continual increase in peak KAM throughout the prolonged walking task. Contrary to our hypothesis, the current participants only exhibited a non-significant 1% (less than 0.01 Nm/kgm) increase in peak KAM across the duration of the walking task (Appendix D). It may be that heavy military-relevant body borne loads and not duration of training, elevate risk of knee OA development for service members. Further considering that individuals present similar non-significant 0% to 5% increases in KAM following general fatigue that is typical of prolonged walking tasks, but significant increases in peak KAM following isolated knee extensor fatigue, future research is needed to determine the specific decrements in muscle function that lead to significant increases in hazardous knee joint moments[37,74,94,95].

In line with existing literature, walking with body borne load did not significantly increase KAA[41,73,96]. The current participants, in fact, only exhibited minimal 0.07º and 0.24º increases in PS KAA with the 15 kg and 30 kg addition of load. Yet, in support of our hypothesis, participants exhibited a 24% and 17% increase in PS and ROM of KAA throughout the prolonged walking task. Substantial increases in knee adduction are reported to load the medial knee joint compartment and may be implicated in OA development[93]. Individuals with OA reportedly exhibit peak KAA angles between 3º and 7º greater and a ROM KAA that is more than 2º greater than healthy controls[23,72]. Although significant, the current participants only exhibited increases in ROM KAA of approximately 0.5º throughout the prolonged walking task, which may be attributed to the concurrent 0.7º increase in PS KAA (Appendix D). Considering the current participants' PS KAA was nearly 4º towards the end of the 60-minute walking task, which puts their knee adduction in line with individuals that have radiographically confirmed OA[97], additional research to determine whether longer walking times and/or distances during load carriage further increase knee adduction angles is warranted.

Contrary to our hypothesis, neither the addition of body borne load nor duration of walking impacted the variability of knee adduction biomechanics. Sufficient variability of knee biomechanics is essential for adequate joint stability, and considered a mechanism to reduce musculoskeletal injury risk[75,98]. Although it was not statistically significant, participants decreased variability of KAA ROM approximately 20% and 30% with the addition of body borne load and duration of walking. This substantial, but not statistically significant, reduction in variability may increase musculoskeletal injury risk. Specifically, large decreases in variability may constrain the joint-level response as well as neuromuscular function around the knee. This may impair the individual's ability to adequately attenuate impact forces, of walking with heavy loads, and increase their injury risk[98–101]. Variability reportedly differs with cadence[20,75]. Considering the current participants may have had natural stride-to-stride variation constrained by stepping to a pre-determined cadence during the walking task, further study is warranted to examine whether this impeded knee adduction variability that would have otherwise been present.

The chosen task may be a limitation. Although the prolonged walking task required participants to walk 1.3 m/s for an hour (or just shy of three miles), the total distance and walking time may have been insufficient to accurately replicate the muscular weakness routinely encountered during military training. However, Lidstone reported substantial biomechanical changes, such as increased vertical ground reaction forces and trunk lean, during a one-hour prolonged load carriage task and thus, we are confident that the chosen one-hour walking time was adequate to produce muscle weakness that may lead to changes in knee biomechanics[45]. Similarly, participants self-reported their ability to safely carry up to 75 pounds but were not required to have prior load carriage experience. Participants with previous load carriage experience may present different knee biomechanics when carrying heavy, military-relevant body borne loads, and warrants further study.

Finally, this study may be limited by the ecological validity, as chosen load carriage equipment or walking courses may not accurately represent military activities. The body borne load was currently applied via a weighted vest, which does not accurately represent the rucksack, body armor, and ammo panel that commonly comprise the load during military activities. In addition, during military activities, service members commonly traverse a variety of terrain, which are typically less uniform than the current walking course.

Conclusion

In conclusion, prolonged walking with heavy body borne load increased knee adduction biomechanics related to knee OA pathogenesis. During the walking task, adding heavy body borne load resulted in a significant increase in peak knee adduction moment, which may load the medial knee joint compartment, and increase knee OA risk. Increased duration of walking lead to greater knee adduction angle, but not moment. The larger knee adduction angles exhibited towards the end of the prolonged walking task may also increase loading of the medial knee joint compartment and risk of knee OA. Neither body borne load, nor time, led to significant changes in knee adduction variability. Yet, participants exhibited a substantial, albeit insignificant, decrease in variability of knee adduction motion with the addition of body borne load and duration of walking. This decreased variability may impact the individual’s ability to attenuate impact forces and disperse joint loading, thereby increasing their injury risk.

CHAPTER FOUR: CONCLUSION

Introduction

The purpose of this study was to: (1) determine whether military-relevant body borne load or (2) the duration of a prolonged walking task led to significant increases in magnitude and variability of knee adduction biomechanics. Key findings support the hypotheses that addition of body borne load and duration of walking results in significant increases in peak knee adduction angle and moment, specific joint biomechanics reported to increase the risk of musculoskeletal injury and disease.

Key Findings

Prolonged walking with heavy body borne load produced a significant increase in knee adduction biomechanics. Specifically, the addition of body borne load produced a significant increase in the magnitude of knee adduction moment, while peak and range of knee adduction angle exhibited a significant increase as duration of walking progressed. These increases in knee adduction biomechanics are reported to load the medial knee joint compartment and may accelerate the wear and tear of the joint's articular surface that characterize OA risk. The variability of both knee adduction moment and angle exhibited no significant changes with the addition of body borne load or duration of walking. Yet, there was a substantial, albeit insignificant, reduction in variability for range of knee adduction motion as body borne load and duration of walking increased. Reduced variability may impede the joint's ability to adequately attenuate the elevated joint loads evident when walking with body borne load and increase musculoskeletal injury risk.

Significance

These findings support the tenet that knee adduction biomechanics exhibited during prolonged load carriage increase musculoskeletal injury and disease risk, particularly knee OA development. Specifically, this study documented that addition of body increased peak knee adduction joint moment, but not peak knee adduction angle; whereas longer duration of walking with body borne load led to greater knee adduction motion. These findings can be used by the military to reduce a service member's risk of musculoskeletal injury and disease development. Specifically, these outcomes can be implemented by the military to identify service members with elevated risk of suffering a training-related musculoskeletal disease as well as improve current military injury prevention and training programs to reduce service members risk of musculoskeletal injury. Successful implementation of the knowledge provided herein by the military may result in a substantial reduction in the number of service members that suffer training-related musculoskeletal injury and decrease the $700 million annually spent by the military treating these debilitating musculoskeletal issues[102].

Limitations

The chosen task and participants may be a limitation. Although the prolonged walking task required participants to walk 1.3 m/s for an hour (just shy or three miles), the duration of the prolonged load carriage task may have been insufficient to accurately replicate the muscular weakness encountered during military training and operations. Previously, however, walking for an hour with body borne load reportedly produced significant biomechanical changes[45], such as increased ground reaction forces and trunk lean. While using a longer duration may have shown more pronounced effects on participant's knee biomechanics, we are confident that the chosen duration is sufficient to elicit changes in knee adduction that occur with body borne load and walking for a prolonged time period. In addition, while each participant self-reported the ability to safely carry up to 75 pounds, they were not required to have prior load carriage experience. Participants with load carriage experience may exhibit different knee biomechanical changes during a prolonged carriage task, but to date, we are unaware of differences in knee biomechanics exhibited between experienced and inexperienced load carriers. Moreover, military recruits typically have little prior load carriage experience before entering basic training, where they commonly suffer knee musculoskeletal injuries.

Finally, this study may be limited by the ecological validity, as chosen load carriage equipment or walking courses may not accurately represent military activities. The body borne load was currently applied via a weighted vest, which does not accurately represent the rucksack, body armor, and ammo panel that commonly comprise the load during military activities. In addition, during military activities, service members commonly traverse a variety of terrain, which are typically less uniform than the current walking course.

Future Work

Prolonged walking with heavy body borne loads altered knee adduction biomechanics. As such, future research is warranted to determine if larger loads and/or longer walking duration, results in further increases of knee adduction biomechanics and injury risk. Moreover, although preliminary analysis was not significant, future work is warranted to determine how static lower limb alignment and strength impact knee adduction biomechanics (Appendices E and F).

Replicating the current work with participants who have prior load carriage experience is also warranted. This might provide additional insight into how experienced service members adapt to prolonged load carriage, and the explicit neuromuscular strategies to target with training protocols to reduce injury risk of such tasks.

Finally, considering service members routinely traverse various terrains (such as deserts, forests and mountains) during training and occupational activities, future study is needed to determine whether they exhibit different knee biomechanical adaptations with changes in terrain, particularly when load or duration of walking increase.

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APPENDIX A

Physical Activity Rating Questionnaire (PAR-Q)

In the table below, write down the number of times (on each day) that you participated in vigorous and moderate physical activities over the last seven days. Examples of vigorous activities would be running, playing sport and training for sport. Examples of moderate activities would be walking or slow cycling. Only include activities if they were undertaken continuously for at least 20 minutes.

	Day 1	Day 2	Day 3	Day 4	Day 5	Day 6	Day 7
Vigorous Activity							
Moderate Activity							

Key:

Physical Activity Score (PAS) = average frequency x 20 x 4 (moderate) + average frequency x 20 x 7.5 (vigorous).

Scoring Criteria:

Low: PAS < 400

Moderate: 400 ≤ PAS < 560

High: PAS ≥ 560

APPENDIX B

Pre-participation Questionnaire

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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 C

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 D

Table D.1. Mean (SD) Peak Stance Knee Adduction Moment, Angle; Initial Contact Angle and Range of Knee Adduction Exhibited with Duration of Walking with Body Borne Load

		0 Min	15 Min	30 Min	45 Min	60 Min
KAM [b] **(N.kg/m)**	**0 kg**	-0.35 (0.07)	-0.35 (0.08)	-0.36 (.08)	-0.36 (0.08)	-0.35 (0.08)
	15 kg	-0.40 (0.09)	-0.40 (0.13)	-0.40 (.10)	-0.41 (0.12)	-0.40 (0.12)
	30 kg	-0.47 (0.10)	-0.49 (0.12)	-0.48 (.12)	-0.49 (0.12)	-0.48 (0.13)
KAA (degrees)	**0 kg**	2.87 (1.98)	3.16 (1.75)	3.48 (1.87)	3.54 (2.11)	3.53 (1.81)
	15 kg	3.12 (1.92)	3.64 (2.37)	3.51 (2.33)	3.82 (2.40)	3.62 (2.42)
	30 kg	2.96 (1.58)	3.50 (1.86)	3.69 (1.85)	3.68 (1.88)	3.94 (1.85)
IC KAA [a] **(degrees)**	**0 kg**	-0.78 (0.97)	-0.58 (1.0)	-0.703 (0.88)	-0.67 (0.81)	-0.65 (0.80)
	15 kg	-0.69 (0.81)	-0.38 (0.89)	-0.430 (0.90)	-0.18 (1.2)	-0.37 (1.1)
	30 kg	-0.73 (0.98)	-0.38 (1.1)	-0.467 (1.0)	-0.51 (1.3)	-0.51 (1.0)
ROM KAA [a] **(degrees)**	**0 kg**	3.64 (2.02)	3.74 (1.78)	4.18 (1.85)	4.21 (2.21)	4.18 (1.81)
	15 kg	3.81 (1.52)	4.02 (2.04)	3.94 (1.94)	4.00 (2.06)	3.69 (1.89)
	30 kg	3.69 (1.42)	3.88 (1.58)	4.16 (1.65)	4.18 (1.76)	4.45 (1.47)

[a] Denotes a significant main effect of time.
[b] Denotes a significant main effect of load

APPENDIX E

Static Knee Adduction Alignment

Static knee adduction alignment was calculated for each participant according to previous literature[103] and submitted to analysis to determine if it confounded knee biomechanics exhibited during the prolonged load carriage task.

For analysis, peak stance KAM and KAA were submitted to an ANCOVA with static knee alignment as a covariate to determine whether alignment impacted differences due to load (*0, 15 and 30 kg)* and time (*0, 15, 30, 45 and 60 min)*

Results

Static knee alignment was -3.64 ± 0.81º. Static alignment was neither a significant covariate for peak stance KAM nor KAA (all: $p>0.05$).

APPENDIX F

Participant Strength

Maximum knee flexion and extension strength calculated during the isokinetic testing were submitted to analysis to determine whether they confound knee biomechanics during the prolonged load carriage task.

For analysis, peak stance KAA and KAM were submitted to an ANCOVA with knee flexor and extensor strength as covariates to determine whether strength impacted differences due to load (*0, 15 and 30 kg)* and time (*0, 15, 30, 45 and 60 min)*.

Results

Peak knee flexion and extension were 85.5 ± 26.5 Nm and 112.5 ± 38.6 Nm respectively. Neither measure of knee strength was a significant covariate for peak stance KAA or KAM (all: $p>0.05$).