PROLONGED LOAD CARRIAGE IMPACTS MAGNITUDE AND VELOCITY

OF KNEE ADDUCTION BIOMECHANICS

by

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ABSTRACT

Introduction: Adopting knee adduction biomechanics during prolonged load carriage, a common military occupational activity, may increase service members knee osteoarthritis (OA) risk. Although service members reportedly increase knee adduction motions and moments during prolonged load carriage, it is unknown if either body borne load or walk duration increases velocity of knee adduction biomechanics, and subsequent knee OA risk. Varus thrust and alignment are also related to greater knee OA risk, yet it is unknown whether varus thrust and/or alignment are related to magnitude and velocity of knee adduction biomechanics during prolonged load carriage. **Purpose:** To determine whether body borne load and walk duration impacted magnitude and velocity of knee adduction biomechanics, or whether increases in knee adduction biomechanics are related to knee varus thrust or alignment. **Methods:** Seventeen participants (11 male/6 female, 23.2 ± 2.9 yrs, 1.8 ± .09 m, 71.0 ± 12.1 kg) had knee adduction biomechanics quantified while walking 1.3 m/s for 60 minutes with three body borne loads (0 kg, 15 kg, and 30 kg). Specifically, peak, average and maximum velocity, as well as time to peak, for knee adduction angle and moment, and varus thrust (first 16% of stance) were calculated at minutes 0, 30, and 60 of the load carriage task. Static knee alignment was calculated as the frontal plane knee projection angle. **Statistical Analysis:** Participants were defined as varus thrust (VT, n=8) or control (CON, n=9). Then, each knee adduction measurement was submitted to a repeated measures ANCOVA to test the main effect and interaction between body borne load (*0 kg, 15 kg, and 30 kg*), time (*minutes 0, 30, and 60*), and group (*VT and CON*), with static alignment considered a covariate. **Results:** A significant 3-way interaction for maximum varus thrust velocity ($p=0.014$), revealed the VT group exhibited greater maximum velocity at minutes 0 through 60 ($p\leq 0.038$) with the 0 kg load, and minutes 0 and 60 ($p\leq 0.043$) with the 15 kg load. Significant load by group interactions for magnitude ($p=0.008$) and average velocity ($p=0.013$) of varus thrust, and maximum KAA velocity ($p=0.041$) revealed VT participants exhibited larger and faster varus thrust and knee adduction angle than the CON group with the 0 kg and 15 kg loads ($p<0.050$). Additionally, both magnitude and maximum velocity of KAM increased with the addition of load ($p=0.009$ and $p=0.004$), and walk duration increased magnitude of varus thrust ($p=0.044$). Static alignment was not a significant covariate for any knee adduction measure ($p>0.05$). **Conclusion:** During prolonged load carriage participants adopted larger, faster knee adduction biomechanics, potentially increasing risk of knee OA. The VT group exhibited greater knee OA risk, and larger, faster knee adduction motions when walking with the lighter (0 kg and 15 kg) loads; while CON adopted increases in knee adduction biomechanics related to knee OA with the heavy (30 kg) load.

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KAM	Knee Adduction Moment
KAA	Knee Adduction Angle
OA	Osteoarthritis
GRF	Ground Reaction Force
VT	Varus Thrust Participants
CON	Control Participants

CHAPTER ONE: INTRODUCTION

Lower limb musculoskeletal disease, such as osteoarthritis (OA), is an ever-increasing problem in the military. Every year over 10,000 service members are diagnosed with lower limb OA, costing upwards of $60 billion dollars to treat[1,2]. The knee joint is the most common location for OA in military populations, and reportedly 100% of service members who suffer occupational knee injury go on to develop OA at the joint[3]. Service members, in fact, are twice as likely to develop knee OA than the general population and the rate among service members steadily rose 45% between 2005 and 2014[2,4]. Knee OA development typically causes loss of joint function and an increase of pain that leads to long term disability and medical discharge[3,5], resulting in a significant occupational burden for the military in general and service members specifically[5]. Considering service member knee OA development may be attributed to routine physical activity with heavy borne loads[6–8], a common military occupational activity, it is imperative to understand the explicit lower limb biomechanics during these activities that increase risk of knee OA development.

Service member knee OA development may be attributed to the changes in lower limb biomechanics that result from heavy body borne loads. Typically, these loads are between 20 kg and 40 kg[6,9], and are carried during all occupational activities, such as locomotion[7,10]. During locomotion, the addition of body borne load leads to significant increases in peak vertical ground reaction forces (GRF). Larger GRFs requires greater force production from the lower limb musculature to prevent lower limb collapse[11], but coincides with a significant increase in limb stiffness. The stiffer limb may transmit greater forces to the soft-tissue structures of the lower limb in general and the knee joint specifically[11–13], increasing the likelihood of soft-tissue injury[14,15]. Additionally, individuals typically exhibit significant alterations in lower limb biomechanics, particularly at the knee, in response to the addition of heavy body borne load[16–18], which may further elevate the forces transmitted to the soft-tissue structures[7]. When running and walking with heavy body borne loads, individuals exhibit significant increases in knee flexion and adduction joint motions and moments[11,19,20]. Of particular importance, are increases in the magnitude of knee adduction biomechanics, involving greater lateral movement of the knee. As the knee moves laterally forces are unevenly distributed through the knee, intensifying the risk for injury and OA development[21–29]. Specifically, knee adduction angle and moment, and varus thrust (rapid lateral motion – i.e., adduction following heelstike[30]) have been directly implicated in the pathogenesis of knee OA[21–29] and reported to increase while carrying heavy body borne loads[19,31–33].

Knee OA is characterized by the degeneration of the joint's articular cartilage that may occur when abnormal forces are placed on the knee[34,35]. The adoption of larger peak knee adduction joint angle and moment, and varus thrust while performing locomotion tasks load may increase the transmission of force to knee joint and associated soft-tissue structures[12,32,36], escalating the risk for knee injury and OA development[21–29,37]. Knee adduction acts to push the knee into varus increasing the peak knee adduction moment, a reported correlate of medial compartment joint loading[38]. Typically, individuals with knee OA exhibit greater amounts of knee adduction moment than individuals without OA, and each 1% increase in knee adduction moment is purported to lead to 6.5 times faster progression of disease at the knee[36]. Individuals that use greater knee adduction during locomotion are reportedly more likely to exhibit varus thrust[30]. Varus thrust is a knee biomechanical parameter thought to indicate joint instability and may represent greater reliance on the knee's passive soft-tissue structures to safely mitigate the impact forces of locomotion[39]. In fact, individuals with visually confirmed varus thrust (>2.5 degrees[30]) during unloaded walking are 4 times more likely to develop knee OA[28]. In addition to magnitude, the velocity of knee adduction biomechanics, as it encompasses both direction and speed of motion[30], may provide greater insight on the transmission of forces to the medial knee joint compartment and risk of OA development. During unloaded walking, Chang et al. reported a significant linear relationship between visualized varus thrust and both magnitude and velocity of knee adduction[30]. Yet it is currently unclear whether walking with heavy body borne load, particularly for extended periods of time, further increases magnitude and velocity of knee adduction biomechanics related to OA development.

Service members are often required to perform occupational-related locomotor tasks for extended periods of time, which may further elevate injury risk[6]. During prolonged bouts of locomotion (i.e., 60 minutes or longer) with body borne load, individuals are reported to increase peak vertical GRF every 15 minutes[22]. This continual increase in GRF may require a concomitant rise in muscular effort to stabilize the knee joint[40], and lead to fatigue induced muscular weakness, resulting in lower limb biomechanics alterations[21–23]. Specifically, during prolonged periods of walking with body borne load the magnitude of knee flexion and adduction motions and moment are reported to increase[20,41]. During a recent prolonged load carriage task, individuals exhibited a significant increase in peak knee adduction moment and angle with 15 kg and 30 kg additions of body borne load and after 30 minutes of walking, respectively[20].

Static knee malalignment has also been identified as a risk factor for knee OA development[4] and may be a precursor to the adoption of hazardous knee adduction biomechanics, especially varus thrust. Individuals that present static knee malalignment, particularly greater varus alignment, reportedly increase risk of knee OA development 2-fold[18]. Varus knee alignment is related to larger peak knee adduction moments[42,43], and may be associated with greater cartilage loss in the knee[29], increasing risk of knee OA development. In addition, when performing unloaded walking individuals who exhibit greater amounts of static knee varus alignment are significantly associated with larger amounts of varus thrust[43], potentially leading to greater instability at the knee and again, increased risk for knee OA development. During loaded locomotion, however, individuals with varus thrust at baseline decrease the magnitude of knee adduction biomechanics related to knee OA, while individuals without varus thrust increase them[44]. It is currently unknown whether individuals with varus thrust exhibit larger increases in magnitude and velocity of knee adduction biomechanics with the addition of heavy body borne load or as duration of walking increases; or whether static knee malalignment is associated with hazardous alterations in knee adduction biomechanics during prolonged load carriage. With that in mind, this study sought to determine whether body borne load and duration of walking impacted magnitude and velocity of knee adduction of knee adduction biomechanics for individuals with and without varus thrust, or whether increases in knee adduction biomechanics during prolonged load carriage are related to static knee varus malalignment.

Specific Aims

Specific Aim 1

To examine the magnitude and velocity of knee adduction biomechanics during a prolonged load carriage task. Specifically, this study seeks to quantify the magnitude and rate of change (velocity) in knee adduction angle and moment, and varus thrust while participants with and without (control) varus thrust walk over-ground at 1.3 m/s for 60 minutes with three different body borne loads (*0 kg, 15 kg, 30 kg*).

Hypothesis 1.1

Participants with varus thrust will exhibit significantly greater magnitude and velocity of knee joint adduction angle, knee joint adduction moment, and varus thrust than the control participants.

Hypothesis 1.2

The addition of body borne load will lead to a significant increase in the magnitude and velocity of knee joint adduction angle, knee joint adduction moment, and varus thrust for all participants, but the varus thrust group will exhibit greater increases than the control group participants.

Hypothesis 1.3

As duration of walking increases there will be significantly greater magnitude and velocity of knee joint adduction angle, knee joint adduction moment, and varus thrust for all participants, but the varus thrust group will exhibit greater increases than the control group participants.

Significance 1:

Determining whether hazardous between knee adduction biomechanics increase with body borne load and/or duration, or whether individuals that present varus thrust exhibit greater changes in knee adduction may provide the military the knowledge to reduce rate of service member knee OA development, as well as knowledge to identify service members at risk of knee OA development. This knowledge can lead to a substantial reduction in healthcare costs associated with treatment of this debilitating disease.

Specific Aim 2

To examine whether the amount of static knee malalignment is related to hazardous knee adduction biomechanics. Specifically, this study seeks to quantify whether greater static knee varus alignment exhibits a significant relation to increase in the magnitude and velocity of knee adduction joint angles and moments, and varus thrust while participants walk over-ground at 1.3 m/s for 60 minutes with three different body borne loads *(0 kg, 15 kg, and 30 kg)*.

Hypothesis 2.1

Static knee varus alignment will exhibit a significant positive relationship to magnitude and velocity of knee joint adduction angle, knee joint adduction moment, and varus thrust.

Hypothesis 2.2

With the addition of body borne load, participants with greater static knee varus alignment will exhibit significantly greater magnitude and velocity of knee adduction joint angle, knee joint adduction moment, and varus thrust than participants without static knee varus alignment.

Hypothesis 2.3

As duration of walking increases, participants with greater static knee varus alignment will exhibit significantly greater magnitude and velocity of knee adduction joint angle, knee joint adduction moment, and varus thrust than participants without static knee varus alignment.

Significance 2

Determining whether static knee varus malalignment is related to the magnitude and velocity of knee adduction biomechanics with body borne load and/ or duration walking may aid the military in identifying service members at risk of knee OA development and will inform training protocols to reduce knee OA development for high risk individuals.

CHAPTER TWO: LITERATURE REVIEW

The following section aims to detail load carriage in the military, specifically; 1) common loads and activities performed with load, 2) musculoskeletal injuries related to load carriage, 3) lower limb biomechanics related to musculoskeletal disease, specifically osteoarthritis, 4) body borne load's effect on these lower limb biomechanics, and 5) the effect prolonged load carriage has on these lower limb biomechanics.

Load Carriage

In the Military

Body borne load carriage is defined as supporting an external mass on an individuals' body (i.e., on the torso), and is a common occupational and recreational activity[8,45]. For example, service members are required to support body borne loads, containing necessary equipment for warfare and survival. In addition to carrying heavy body borne loads during occupational activities, many training exercises (i.e., marching, running, hiking, and walking) are performed with similar body borne loads[2]. On average these body borne loads range from 20 kg to 40 kg, but can reach 68 kg during certain activities[6,9,17,46], exceeding recommendations set by the military[47,48]. Service members also locomote for prolonged periods of time, upwards of 20 km a day with heavy body borne loads[6], causing repetitive overloading of lower limb soft tissue and bone, which can be detrimental to service members' long term health[49]. Taking this into consideration, it is not a surprise that service members are at increased risk for lower limb musculoskeletal injury and disease[7,9].

Musculoskeletal Injury

Injury in the Military

Musculoskeletal injuries often stem from damage occurring at the soft tissue and bone caused by the physically demanding nature of the military[49,50], and can occur in the muscles, nerves, tendon, joints, cartilage, and spinal discs. In the military musculoskeletal injuries are a major occupational burden, and overuse has been found to be the most prevalent mechanism[49]. As previously mentioned, it is not uncommon for service members to hike and walk up to 20 km a day with heavy body borne loads[6], which places abnormal forces on joints, increasing the risk for injury[8,16].

Incidence in the Military

Of all injuries among service members, 55% are musculoskeletal injuries[51,52]. During basic training 19% to 40% men, and 40 to 70% of women, were estimated to sustain a musculoskeletal injury[46,53], resulting in up to 30% of all service members not being deployable[45]. Disability rates in the Army alone have increased 6-fold since 1980, mainly attributed to musculoskeletal injuries that occur due to load carriage[17]. As a result of service members being disabled from musculoskeletal injury, healthcare costs and lost wages also increase[49,50]. In 2012 alone, more than $700 million was spent on the treatment of musculoskeletal injuries[45], and in 2018 nearly $6 billion was lost in service member wages[54].

Approximately 62% of musculoskeletal injuries are reportedly caused by marching with heavy body borne loads[55]. Previous literature has suggested that supporting more than 30 kg of load while locomoting reportedly increased risk for musculoskeletal injury by more than 100%[55]. The most common site for a musculoskeletal injury is within the lower limbs, more specifically greater than 80% of all musculoskeletal injuries occur in the lower limbs[49,50,56]. Within the lower limbs, the knee makes up approximately half of all non-combat musculoskeletal injuries[55,57,58]. Unfortunately, when a service member experiences one knee injury they are significantly more likely to develop repeat injuries, which, long term leads to the development of musculoskeletal diseases, such as osteoarthritis (OA)[1,2]. In fact, 100% of service members that suffer from a knee injury develop knee OA[3].

Musculoskeletal Disease

Osteoarthritis in the Military

Osteoarthritis is characterized by the degeneration of articular cartilage associated with abnormal loads placed on the knee joint, often accompanied by pain and results in loss of joint function[2,35,59]. Previous injury and heavy body borne loads have been identified as risk factors for lower limb OA development[6,7,19], and in recent years the incident rate of knee OA has significantly increased, especially in populations that are routinely physically active while carrying body borne loads, such as the military[2,4]. Every year an average of 10,287 active duty service members are diagnosed with OA, resulting in more than $60 billion dollars spent on treatment[1].

Knee OA Biomechanics

A common location of OA among service members is the knee joint, and in recent years the incident rate has been on the rise[2,4]. Between 2005 and 2014 knee OA rates have increased by 45%[2,4] and when compared to the general population, service members experience knee OA at twice the rate[4]. Again, this is largely due to the fact that the military occupational tasks are very physically demanding and heavy body borne loads are involved. Knee OA development can be attributed to abnormal loads placed on the knee joint due to changes in lower limb biomechanical variables present in walking[2,35]. During walking it is common to see vertical ground reaction forces within the medial compartment of the knee, where knee OA is most common, increase 2.5 times, creating a greater likelihood of developing knee OA [37]. Furthermore, there is a direct relationship between increased vertical ground reaction force and greater knee joint loading[21,39]. Common measures that have been associated with knee joint loading include dynamic knee adduction biomechanics, as well as static alignment of the knee. Specifically, peak knee adduction angle and moment, varus thrust, and static knee malalignment have all been reported to increase the odds of knee OA development and progression[21–23,25], with knee adduction moment being considered a good clinical measure of medial compartment loading[38].

Dynamic Knee OA Biomechanics

Increases in peak knee adduction angle and moment, and varus thrust have been associated with increased risk for knee OA development and progression[21–23], all of which represent greater lateral movement of the knee creating uneven knee joint loading[37]. The external knee adduction moment (KAM) is a common measure that correlates with knee joint loading, and can be used as a clinical surrogate for medial compartment loading[38]. KAM is defined as the ground reaction force vector passing medial to the knee, and is a strong predictor of disease severity as well as presence of symptoms, with significantly higher peak KAM values occurring in affected individuals[60,61]. In fact, it has been observed that for every 1% increase in KAM, the progression of knee OA increases 6.5 times[36]. Knee adduction angle (KAA) during gait is another measure that shows an association with knee OA progression, with higher KAA values being observed in affected subjects [62]. KAA during dynamic trials can also be looked at as an increase in knee varus, which has also been shown to increase knee OA progression rate[26]. Greater KAA during gait increases the amount of bone on bone contact in the medial compartment, increasing wear and tear on the articular cartilage[63,64]. In addition, increases in KAA during walking push the into greater varus, increasing the peak knee adduction moment and varus thrust. Varus thrust, or the abrupt increase in KAA during the initial stages of stance[30], is another variable that has been looked at in the progression of knee OA. During locomotion this abrupt change in KAA has been identified as a potential risk factor for knee OA progression[28]. When varus thrust has been observed there is a 4-fold increase in knee OA incidence[28], KAA peak increases[30], and KAM peak is significantly greater than those who do not present varus thrust [43]. Although not observed as much as peak values, the velocity that knee adduction biomechanics occur may provide additional insight on forces placed on the knee joint since it encompasses both direction and speed of the movement[30]. One instance where knee adduction angular velocity was observed, greater peak knee adduction angle and moment, and varus thrust was associated with increases in velocity[30], all implicated it the risk and progression of knee OA[24–29].

Static Alignment and Knee OA

Static malalignment at the knee has been associated with knee OA development, with varus and valgus alignment increasing OA progression by 4 and 5 fold, respectively[65]. More recently varus alignment, but not valgus alignment of the knee joint, has been shown to increase the risk of OA development by 2-fold[18,25], as varus alignment is associated with greater amounts of cartilage loss in the knee[29]. This is mainly due to the fact that subjects who present varus alignment in the knee regularly increase the loading on the medial compartment of the knee[71] in the form of larger knee adduction moments[42]. In fact, it has been reported that varus static alignment in patients with knee OA was the best single predictor of peak external knee adduction moment[66], and as stated before, greater peak knee adduction moments have been associated with faster progression and likelihood of knee OA[63,64]. In addition, an increase in varus alignment at baseline is also associated with a greater level of knee OA severity[67], creating a greater loss of joint function and increased joint pain.

Effect of Body Borne Load

Physiological

Biomechanical and physiological parameters of locomotion change with excess weight, no matter the form of that weight (i.e., backpack, rucksack, etc.). Physiologically, walking with heavy body borne loads causes an increase in oxygen uptake, metabolic cost, work intensity, heart rate, and ventilation[68]. Walking with a backpack, compared to no backpack, produces 30% to 45% higher energy expenditure, starting with loads that are 15% of person's body mass[69,70]. Generally, as loads increase the energy expenditure increases proportionally, but also depends on the position of the load and speed of locomotion[69,70]. When the load is located closer to the center of mass and higher up on the back, the metabolic cost of load carriage decreases compared to alternative methods of load carriage[7,9,70].

Spatiotemporal

Altered lower limb biomechanics have also resulted from heavy body borne load, as subjects try to compensate for added load[32,33], adding to the risk of musculoskeletal injury and disease. These spatiotemporal changes have been observed with loads as low as 8 kg[71], and are characterized by changes in the gait cycle, specifically the double (both feet on the ground) and single (one leg swinging through the air) support phases of walking. The addition of body borne load increases subjects' time spent in double support, decreases stride length, and increases stride frequency[11,72,73]. Despite this, it is usually found that during a fixed pace, as opposed to a self-selected speed, stride length and frequency change the most[71]. The increased time spent in double support and alterations to stride length and frequency allows individuals to absorb higher ground reaction forces associated with heavy body borne loads[11,74].

Ground Reaction Force

Ground reaction force (GRF) is a common measure examined during load carriage and can provide key insight of gait and impact forces acting upon the lower limbs. The addition of heavy body borne loads produce significantly higher peak vertical and anterior-posterior ground reaction forces have been observed[11], with vertical GRF impact peaks increasing by 5% to 10%, or in some cases increasing proportionally with added load[41,67,73]. Anterior-posterior forces have also been shown to increase proportionally with added load[74]. These elevated GRFs require greater muscle involvement to prevent lower limb collapse, however, this attempt to prevent limb collapse increases the forces placed on the lower limbs in general and on the knee joint specifically[11]. Elevated ground reaction forces have also been reported to decrease the medial-lateral lower limb stability, again increasing the risk for injury and disease[11,74,75].

Trunk and Hip Kinematics

Areas that exhibit biomechanical changes during load carriage are the trunk and hips. Typically, in response to added body borne load, as low as 6kg, subjects increase anterior lean of the trunk and head[17]. This forward tilt of the trunk and head lead to increased muscle activity in the pelvis and lower back in an attempt to gain back postural stability and offset the altered location of center of mass[17]. As body borne load increases during walking, hip range of motion typically increases, but this is not always the case. Birrel[74] and Attwells[71] reported no changes in hip range of motion when comparing 0 kg and 15 kg loads, and 0 kg and 32 kg loads. At initial contact hip angle values increase with the addition of body borne load[76]. Linear increases in hip flexion have been reported with any loads between 7.5 kg and 40kg, but no significant changes with heavier loads[71,72].

Knee Kinematics and Kinetics

In addition to the trunk and hip, biomechanical changes also occur at the knee joint. During the weight acceptance phase of gait, the knee acts as a shock absorber in an attempt to mitigate increased vertical ground reaction force[74], requiring greater force production from the lower limb musculature, increasing lower limb joint stiffness, thereby resulting in a greater reliance on soft-tissue and bone to further absorb the increase in vertical ground reaction force[14,15] . Typically knee flexion range of motion significantly increases between 0 kg loads and loads above 15 kg, however, no change in knee flexion range of motion has previously been observed between 0 kg and 15 kg loads[71,72,74]. At times, there can even be a decrease in knee joint flexion range of motion[71,72], which potentially increases stiffness in the lower limb, another risk factor for musculoskeletal injury and disorder[11–13]. Less information is known about frontal plane knee joint motions (knee adduction) while under load, and current data is less consistent. Birrel[74] found no changes in frontal plane biomechanics during walking with body borne load, while others have seen significant changes in frontal plane biomechanics, mainly during running trials[31–33]. For example, Brown[33] reported that during running knee adduction angle and moment significantly increased with 30% of body weight, but in 2018 reported those with varus thrust at baseline reduced knee adduction biomechanics associated with OA as loads got heavier, and those without varus thrust increased knee adduction biomechanics related to knee OA[44]. Again, although more concrete data is known about knee flexion-extension biomechanics, looking deeper into knee adduction biomechanics may be more beneficial as they directly relate to OA incident rates and rate of progression[63,64].

Ankle Kinematics

The ankle range of motion has been reported to significantly increase during locomotion with body borne loads[12]. Individuals try to increase propulsive forces through greater amounts of ankle plantar flexion, and then have a rapid change back to a dorsiflexed position[12]. Walking speed also has an impact on ankle range of motion with body borne load. The greater the load is the more effort must be put into locomotion to maintain the same speed, possibly explaining why ankle range of motion increases[77]. Ankle dorsiflexion increases seen during body borne loading have been attributed to individuals attempting to increase knee flexion to absorb added forces, as greater amount of dorsiflexion seems to facilitate larger amounts of knee flexion[78]. Conversely, one study observed no significant changes in total ankle range of motion with load, but like others saw an increase in plantar flexion[76].

Prolonged Load Carriage

Muscular Weakness

In the military service members perform high volumes of training, oftentimes involving activities like marching, walking, and hiking for extended periods of time with body borne loads[6]. For example, it is not uncommon for military service members to cover over 20 km a day with heavy body borne loads[6]. During these prolonged bouts of physical activity service members can experience muscle induced weakness within the lower limb, which increases the risk for knee injury[6,40]. Because vertical ground reaction forces increase significantly during locomotion due to heavier body borne load greater muscle involvement is needed to stabilize the lower limb[11–13]. However, this attempt to stabilize the lower limb through greater muscle involvement causes an increased rate of muscle weakness, further increasing vertical ground reaction force and limits the limbs ability to absorb additional force[21–23,40]. For example, Lidstone[22] observed significant increases in vertical ground reaction force every 15 minutes of a 60 minute walking task, reflecting the impact fatigue induced muscle weakness has on the attenuation of force. Again, this places a greater reliance on the passive structures, such as bone and cartilage, increasing the risk for injury and disease[14,15]. More specifically, increased muscle recruitment increases compressive force within the knee joint, causing greater bone on bone contact[19,79]. This combination of greater muscle force and higher vertical ground reaction force may lead to further alterations in lower limb biomechanics related to knee musculoskeletal injury and disease, especially when walking for long periods of time[21–23]. With the attempt to absorb force comes greater knee flexion range of motion[72]. In addition, peak knee adduction angle and moment, and varus thrust may increase during prolonged walking, adding to the potential risk of musculoskeletal injury and disease of the knee[21–23]. Although sagittal plane motions and ground reaction forces have been observed during prolonged walking[22], there has been less focus placed on frontal plane biomechanics during prolonged walking with body borne, in particular how the velocity and magnitude of knee adduction biomechanics change.

Summary

Musculoskeletal injuries of the lower limb, specifically at the knee joint, significantly increase the risk for musculoskeletal diseases like osteoarthritis. Musculoskeletal injuries and disease are an extreme occupational and financial burden for individuals who participate in intense physical activity while carrying heavy borne loads. An example of this is military service members, who are 2 times more likely to develop knee OA when compared to the general population, largely due to heavy body borne loads and repeat injuries. These heavy body borne loads seen in military occupations increase the magnitude of knee biomechanics related to the progression and development of knee OA, specifically knee adduction angle and moment, and varus thrust. Prolonged intense physical activity also induces changes in these biomechanics related to knee OA through causing fatigue induced muscle weakness. The longer an individual is exercising the less effective they are at mitigating forces seen at the knee, creating a larger reliance on passive structures like bone and cartilage. When heavy body borne loads are combined with prolonged activity, there is a potential compounding effect that further increases the risk for injury and disease development. However, previous literature has looked at differences in knee adduction biomechanics while either locomoting for an extended amount of time or with load, but not together. Also, previous studies have not explored the rate, or velocity, at which these knee adduction biomechanics occur during prolonged load carriage with body borne loads. This would provide supplementary information on the risk factors for knee OA development within military populations.

CHAPTER THREE: MANUSCRIPT

Introduction

Osteoarthritis (OA) is a significant occupational burden for the military in general and service members specifically[5]. Every year over 10,000 service members are diagnosed with lower limb OA, costing upwards of $60 billion dollars to treat[1,2]. The knee joint is the most common location for OA in military populations, and reportedly 100% of service members who suffer occupational knee injury go on to develop OA at the joint[3]. Service members, in fact, are twice as likely to develop knee OA than the general population and the rate among service members steadily rose 45% between 2005 and 2014[2,4]. Knee OA development typically causes loss of joint function and an increase of pain, leading to long term disability and medical discharge for service members[3,5].

Service member knee OA development may be attributed to altered lower limb biomechanics when walking with heavy body borne loads[6–8]. Locomoting with body borne load leads to significant increases in peak vertical ground reaction forces (GRF)[22], and requires greater force production from lower limb musculature to prevent limb collapse[11]. Yet, the larger GRFs and muscle force coincide with a significant increase in limb stiffness[32]. The stiffer limb may transmit greater impact forces to the soft-tissue structures of the lower limb in general and the knee joint specifically[11–13], increasing the likelihood of soft-tissue injury[14,15]. In response to the heavy body borne loads service members reportedly adopt hazardous knee biomechanics[16–18], potentially further elevating the risk of soft-tissue damage and OA development[7]. Of particular importance, are increases in the magnitude of knee adduction biomechanics. Specifically, magnitude of knee adduction angle and moment, and varus thrust (rapid lateral knee motion – i.e., adduction following heelstike[30]) have been directly implicated in the pathogenesis of knee OA[21–29], and are reported to increase when walking with heavy body borne loads[19,31–33].

Knee OA is characterized by the degeneration of the joint's articular cartilage and may occur when abnormal joint forces damage the knee's soft-tissues[34,35]. The adoption of larger peak knee adduction joint angle and moment, and varus thrust when walking with load may increase the transmission of force to knee joint and associated soft-tissue structures[12,32,36], escalating the risk for knee injury and OA development[21–29,37]. Knee adduction acts to push the knee into varus increasing the peak knee adduction moment, a reported correlate of medial compartment joint loading[38]. Typically, individuals with knee OA exhibit greater amounts of knee adduction moment than individuals without OA, and each 1% increase in knee adduction moment is purported to lead to 6.5 times faster progression of disease at the knee[36]. Individuals that use greater knee adduction during locomotion are reportedly more likely to exhibit varus thrust[30]. Varus thrust is a knee biomechanical parameter thought to indicate joint instability and may represent greater reliance on the knee's passive soft-tissue structures to safely mitigate the impact forces of locomotion[39]. In fact, individuals with visually confirmed varus thrust (>2.5 degrees[30]) during unloaded walking are 4 times more likely to develop knee OA[28]. In addition to magnitude, the velocity of knee adduction biomechanics, as it encompasses both direction and speed of motion[30], may provide greater insight on the transmission of forces to the medial knee joint compartment and risk of OA development. During unloaded walking, a significant linear relationship was observed between visualized varus thrust and both magnitude and velocity of knee adduction[30]. Yet it is currently unclear whether walking with heavy body borne load, particularly for extended periods of time, further increases magnitude and velocity of knee adduction biomechanics related to OA development.

Service members are often required to perform occupational-related locomotor tasks, such as walking or marching, for extended periods of time[6]. During prolonged bouts of walking l (i.e., 60 minutes or longer) with body borne load, individuals are reported to increase peak vertical GRF every 15 minutes[22]. This continual increase in GRF may require a concomitant rise in muscular effort to stabilize the knee joint[40], and lead to fatigue induced muscular weakness, resulting in lower limb biomechanics alterations[21–23]. Specifically, during prolonged periods of walking with body borne load, the magnitude of knee flexion and adduction joint angle and moment are reported to increase[20,41]. During a recent prolonged load carriage task, individuals exhibited a significant increase in the magnitude of knee adduction angle and moment after 30 minutes of walking and the addition of 15 kg and 30 kg body borne loads, respectively[20].

Static knee malalignment has also been identified as a risk factor for knee OA development and may be a precursor to the adoption of hazardous knee adduction biomechanics, especially varus thrust[4,30]. Individuals that present greater varus alignment reportedly increase risk of knee OA development 2-fold[18]. Varus knee alignment is associated with larger peak knee adduction moments and greater magnitude of varus thrust during unloaded walking[42,43]. Yet, it is currently unknown whether static knee malalignment is associated with hazardous alterations in knee adduction biomechanics during prolonged load carriage. This study sought to determine whether body borne load and duration of walking impacted magnitude and velocity of knee adduction biomechanics for individuals with and without varus thrust, and whether static knee varus malalignment leads to greater increases in knee adduction biomechanics during prolonged walking. We hypothesized that varus thrust participants would exhibit greater increases in magnitude and velocity of knee adduction biomechanics with the addition of body borne load and walk duration than the control participants, and static knee varus malalignment would exhibit a significant positive relationship with magnitude and velocity of knee adduction biomechanics during a prolonged load carriage task.

Methods

Participants

We recruited 17 participants for this study (Table 3.1). Each participant was between 18 and 40 years of age, recreationally active as defined on the Physical Activity Readiness Questionnaire (Appendix A)[80] and completed a pre-participation questionnaire (Appendix B). To be included potential participants had to self-report the ability to safely walk with 75 pounds. Potential participants were excluded if they reported: 1) history of surgery in the low back or lower extremities; 2) recent (within the last six months) pain and/or injury located in the back or lower extremity; 3) any known neurological disorder; and/or 4) currently pregnant. Prior to testing, research approval was obtained from the local Institutional Review Board, and all participants provided written informed consent.

Table 3.1. Subject Demographics

	N	Age	Height (m)	Weight (kg)
Varus Thrust	8	23 ± 1.9	1.79 ± 0.1	73.1 ± 14.5
Control	9	23 ± 4.1	1.73 ± 0.1	69.3 ± 9.6

Experimental Design

Each participant completed three test sessions. During each test session, participants completed a prolonged walk task with a different body borne load (0 kg, 15 kg, and 30 kg) (Picture 3.1). For each body borne load, participants wore tight fitting spandex shorts and a shirt. For the 15 kg and 30 kg loads, participants also donned a weighted vest (V-MAX, WeightVest.com, Rexburg, ID, USA) that was systematically adjusted to provide the additional load. Prior to testing, the vest weight was determined and only loads within ± 2% of the targeted weight were accepted. Before testing, a 3 x 3 Latin square was used to randomly assign every participant a test order for each load condition (Table 3.2). All test sessions were separated by at least 24 hours to minimize injury risk from fatigue.

Picture 3.1. Spandex and weight vest set-up used.

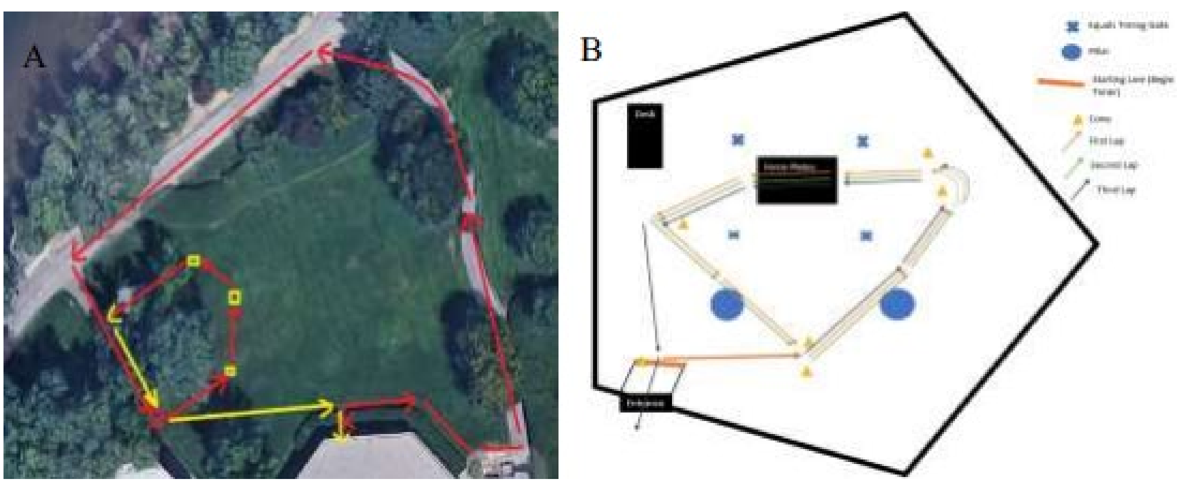
Table 3.2. Latin Square design that will be used to randomize the testing order for each weight condition

	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

Prior to testing, each participant had lower limb (hip, knee, and ankle) strength data recorded on an isokinetic dynamometer (HUMAC NORM, CSMI, Stoughton, MA, USA). To record lower limb strength, each participant performed three maximal isometric contractions with their dominant limb for hip and knee flexion and extension, hip adduction, and ankle plantar and dorsiflexion. For hip flexion and extension contractions, participants stood upright and had their hip secured at 15 degrees of flexion. For the knee flexion and extension contractions, participants were seated on the dynamometer with hip and knee secured at 85 degrees and 60 degrees of flexion to the dynamometer, respectively[81]. For hip adduction contraction, participants had their dominant hip abducted to 15 degrees while laying on their non-dominant side[82,83]. For the ankle plantar and dorsiflexion contractions, participants laid prone on the dynamometer with the ankle in a neutral position (0 degrees of plantar flexion). Then, for each isometric contraction, the participants performed three maximal 5 second isometric contractions, with 40 seconds of rest between each contraction[84]. The maximum torque produced during the three trials was recorded and normalized to the participant's body mass. The participant's dominant leg was determined by simply asking them what foot they would kick a ball with[85].

Biomechanical Testing

During each test session, participants completed the prolonged walk task. The prolonged walk task required participants to walk continuously over-ground at 1.3 m/s for 60 minutes. During the 60-minute walk task, each participant completed 13 laps of a predetermined course that was approximately 390 meter in length and consisted of indoor and outdoor portions (Picture 3.2). Each lap of the walk course required participants to complete one pass through the indoor and outdoor portions every 5 minutes. Participants began the walk task in the laboratory at minute zero, and were required to complete three walk trials through the motion capture volume before proceeding to the outdoor portion. For each walk trial, participants walked 1.3 m/s (± 5%) through the motion capture volume and over a force platform. The speed of each walk trial was recorded with two sets of infrared timing gaits (TracTonix TF100, TracTonix Wireless Timing Systems, Lenexa, KS), placed 4 meters apart in the capture volume. A successful walk trial required participants walk the correct speed and only contact the force platform with their dominant limb. After completion of the walk trials, participants immediately proceeded to the outdoor portion of the course, where they followed a marked route that traveled over asphalt and grass, and returned the participant to the laboratory door. Throughout the walk task, participants were required to step to a metronome, set to a predetermined cadence, to ensure they walked the correct speed for both the indoor and outdoor portions.



Picture 3.2. Outdoor (A) and Indoor (B) portions of the prolonged walking task

<u>Biomechanical Analysis</u>

During each walk trial, participants had three dimensional (3D) lower limb (hip, knee, and ankle) biomechanics recorded. Specifically, eight high speed (240 hz) optical cameras (MXF20, Vicon Motion Systems LTD, Oxford, UK) recorded lower limb motion data, while synchronous ground reaction force (GRF) data was collected with one in-ground force platform (2400 hz, AMTI OR6 Series, Advanced Mechanical Technology Inc., Watertown, MA).

For each walk trial, lower limb biomechanical data was quantified from the 3D coordinates of 34 retroreflective markers and 4 virtual markers (Table 3.3). 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., Germantown, MD). After each marker was placed, participants stood in anatomical position for a static recording that was used to create a kinematic model. The seven-segment kinematic model (pelvis, and bilateral thigh, shank, and foot) was constructed in Visual 3D (v6, C-Motion Inc, Germantown, MD, USA) by assigning a local coordinate system with three orthogonal axes (x, y, and z) to each segment. For the pelvis, the local coordinate system had 3 degrees of rotational and translational freedom, and a joint center was defined as the halfway point between the right and left anterior iliac spine. For the hip, the local coordinate system had 3 degrees of freedom and a functional joint center was determined in accordance with Rozumalski and Schwartz[86]. Both the knee and ankle, were assigned a local coordinate system with three degrees of freedom, and had joint centers defined as the midpoint between medial and lateral femoral epicondyle and medial and lateral malleoli in accordance to Grood and Suntay[87], and Wu[88], respectively.

Table 3.3. Marker placement for the kinematic model.

Body Segment	Markers
Trunk	*xiphoid process, clavicular notch, c7 vertebrae, bottom of the scapula,* **acromion processes**
Pelvis	**anterior-superior iliac spines, posterior-superior iliac spines**, iliac crests
Thigh	greater trochanters, **lateral epicondyles, medial epicondyles**, distal thighs
Shank	tibial tuberosities, lateral fibulas, distal tibias, **lateral malleoli, medial malleoli**
Foot	**first metatarsal heads, fifth metatarsal heads**, heels, midpoint of first and fifth metatarsals

Bold indicates calibration markers, *italics* indicate virtual markers, and the rest are tracking markers.

For each walk trial, the synchronous GRF and marker trajectory data were filtered using a fourth-order Butterworth filter (12Hz), and then, knee biomechanics were calculated in Visual 3D. In Visual 3D, the filtered marker trajectories were processed to calculate knee joint rotations expressed with respect to each participants' static pose using the joint coordinate system approach[87,88], while the filtered kinematic and GRF data were processed to obtain 3D knee forces and moments using standard inverse-dynamics analysis. Segmental inertial properties were defined according to Dempster[89], and the knee moments are expressed as external and normalized to the participants' height (m) and weight (N).

Custom MATLAB (MATLAB r2018a, Mathworks, Natick, MA) code was used to calculate average and maximum velocity of stance phase knee adduction biomechanics. Stance phase was identified as heel strike to toe-off, and defined as the first instance the vertical GRF ascends and descends past 10 N, respectively. Average velocity of knee adduction angle and moment were calculated as the change in angle (or moment) from initial contact to peak value exhibited during stance phase divided by the corresponding change in time from initial contact to peak value. Maximum knee adduction angle and moment velocity was defined as the largest instantaneous velocity exhibited from initial contact to peak angle (or moment) value exhibited during stance. In addition, average and maximum velocity of varus thrust, or the knee adduction angle exhibited during the first 16% of stance, were also calculated[30]. Specifically, the average and maximum varus thrust velocity was defined as the change in knee adduction exhibited during the first 16% of stance divided by the corresponding change in time, and the maximum varus thrust velocity was the largest instantaneous velocity of knee adduction angle during the first 16% of stance.

Participants also had static knee alignment calculated, as the frontal plane knee projection angle (ab-adduction). Specifically, static knee alignment was calculated with the participants standing in anatomical position using hip, knee, and ankle joint centers, according to Mizner et al[90], during the static recording.

Statistical Analysis

For statistical analysis, participants who exhibited knee adduction equal to or greater than 2.5 degrees[30,43], during the first 16% of stance at minute 0 when walking with the 0 kg load, were assigned to the varus thrust group (VT = 8, range = 2.69 to 5.78 degrees), whereas participants who exhibited less than 2.5 degrees of knee adduction were assigned to the control group (CON; N=9, range 0.92 to 2.18 degrees).

Knee adduction biomechanics including, average and maximum velocity for knee adduction angle (KAA) and moment (KAM), and varus thrust, as well the magnitude of and time to peak for KAA, KAM, and varus thrust were submitted to statistical analysis. Each dependent variable was averaged across two walk trials recorded at minutes 0, 30, and 60 of the prolonged walk task, and then submitted to a repeated measures ANCOVA to test the main effect and interaction between body borne load (*0 kg, 15 kg, and 30 kg*), time (*minutes 0, 30, and 60*) and group (*VT and CON*). Static knee alignment was considered a covariate for each ANCOVA. Significant interactions were submitted to a simple effects analysis and a Bonferroni correction was used for significant pairwise comparisons. Alpha was set *a priori* $p<0.05$. All statistical analysis was performed using SPSS software (v25 IMB, Armonk, NY, USA).

Results

A significant 3-way interaction was observed for maximum varus thrust velocity ($p=0.014$) (Figure 3.1). The VT group exhibited greater maximum velocity at minutes 0 ($p=0.004$), 30 ($p=0.007$), and 60 ($p=0.038$) with the 0 kg load, and greater velocity at minutes 0 ($p=0.027$) and 60 ($p=0.043$) with the 15 kg load compared to CON. However, similar group differences were not observed with the 30 kg load ($p>0.05$). Although the VT group did not exhibit significant changes in maximum velocity during the walk task ($p>0.05$), at minutes 60, the CON group exhibited greater maximum velocity with the 30 kg load compared to the 0 kg ($p=0.037$) and 15 kg ($p=0.030$) loads, because, with the 30 kg load, CON increased maximum velocity at minute 60 compared to minute 0 ($p=0.049$).

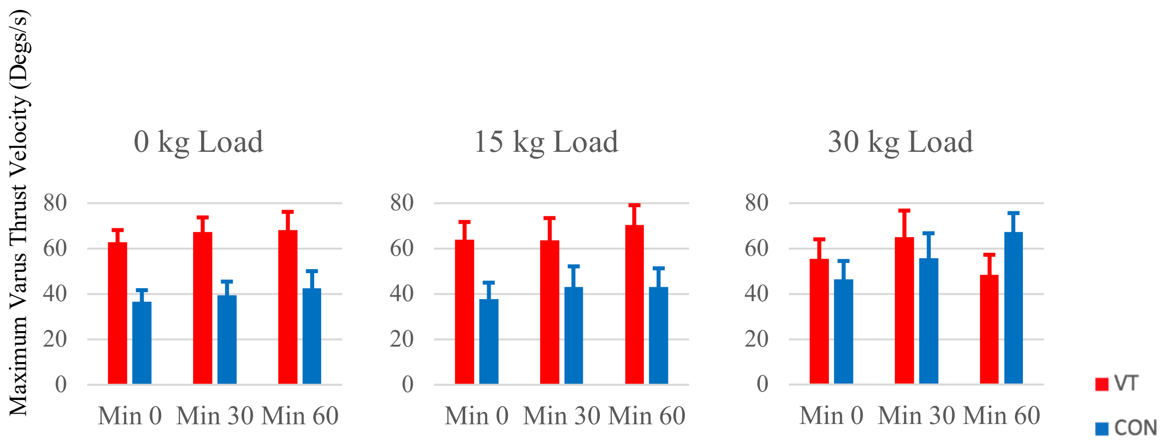


Figure 3.1. Maximum varus thrust velocity for each time point (minutes 0, 30, and 60) during each of the body borne loads (0, 15, and 30 kg).

The ANCOVA revealed significant load by group interaction for magnitude (p=0.008) and average velocity (p=0.013) of varus thrust. Specifically, VT exhibited greater magnitude and velocity of varus thrust than CON participants with the 0 kg (p<0.001 and p<0.001) and 15 kg (p=0.031 and p=0.025) (Figure 3.2) loads, but no group differences were observed with the 30 kg load (p>0.05).

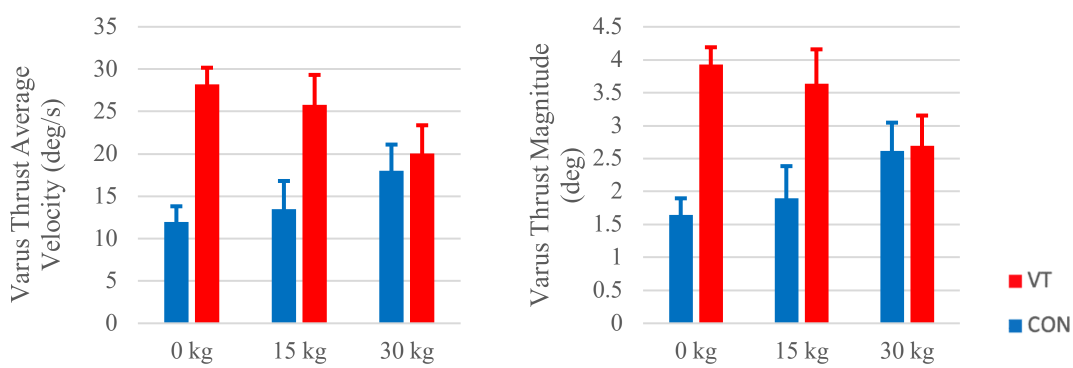


Figure 3.2 Comparison of VT and CON for average varus thrust velocity (A) and magnitude of varus thrust (B) during each body borne load (0, 15, and 30 kg)

A significant load by group interaction was evident for maximum KAA velocity (p=0.041). The VT participants exhibited greater maximum KAA velocity than CON with 0 kg (p=0.011) and 15 kg (p=0.050) loads, but not the 30 kg (p=0.747) load.

A significant time and group interaction was observed for average KAM velocity ($p=0.049$). However, after correcting for Type I error, significant differences between groups and times were not evident ($p>0.05$).

Load

Load had a significant effect on magnitude and velocity (maximum) of KAM ($p=0.009$ and $p=0.004$), but not KAA or varus thrust ($p>0.05$) (Figure 3.3). Both magnitude and maximum KAM velocity were greater with the 15 kg ($p=0.002$ and $p=0.014$) and 30 kg ($p=0.021$ and $p=0.012$) load conditions compared to the 0 kg load condition, but when comparing the 15 kg and 30 kg loads no significant difference in magnitude ($p=0.407$) or maximum KAM velocity ($p=0.384$) was observed. Load had no significant effect on time to peak or average KAM velocity ($p>0.05$).

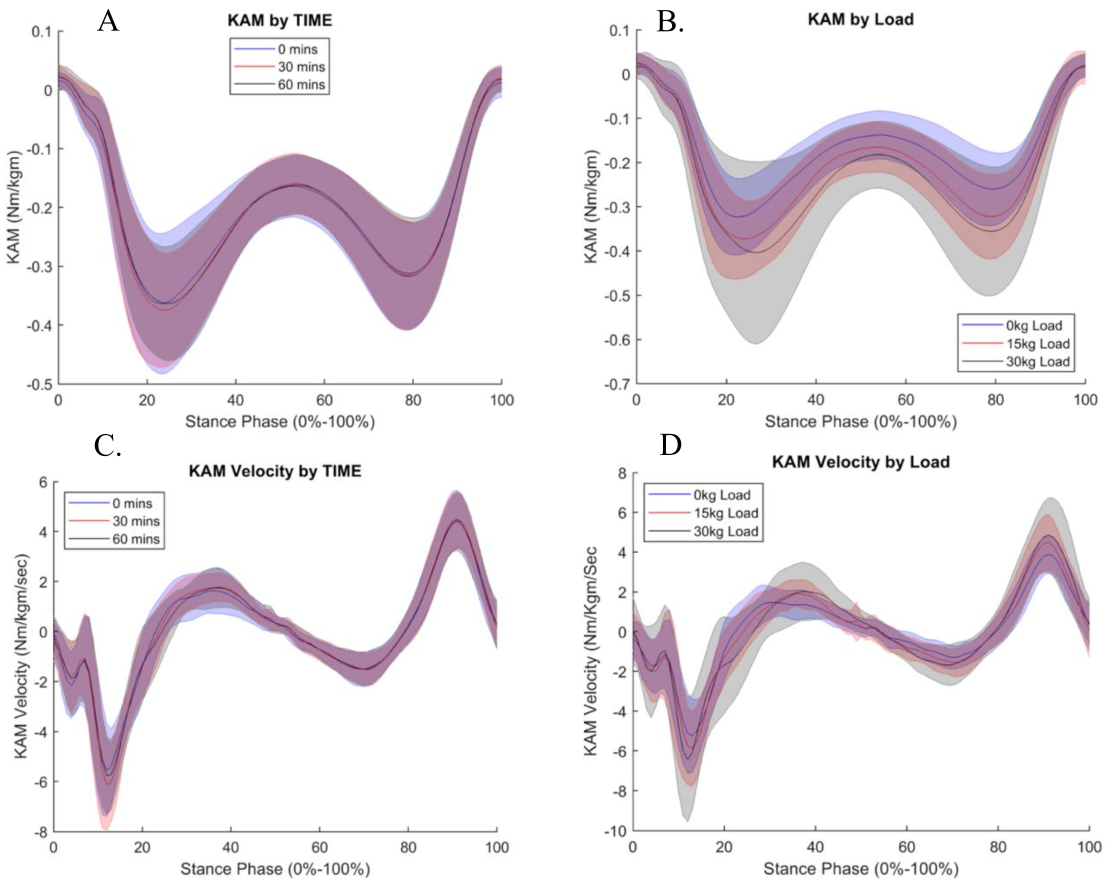


Figure 3.3. Mean ± SD stance phase (0-100%) magnitude of KAM across time (A) and load (B), and velocity of KAM across time (C) and load (D).

<u>Time</u>

Time had a significant effect on magnitude of varus thrust (p=0.044), but no other knee adduction measure (p>0.05) (Figure 3.4). Specifically, magnitude of varus thrust was significantly greater at minutes 30 (p=0.038) and 60 (p=0.050) compared to minute 0, but no difference was evident between minutes 30 and 60 (p>0.999).

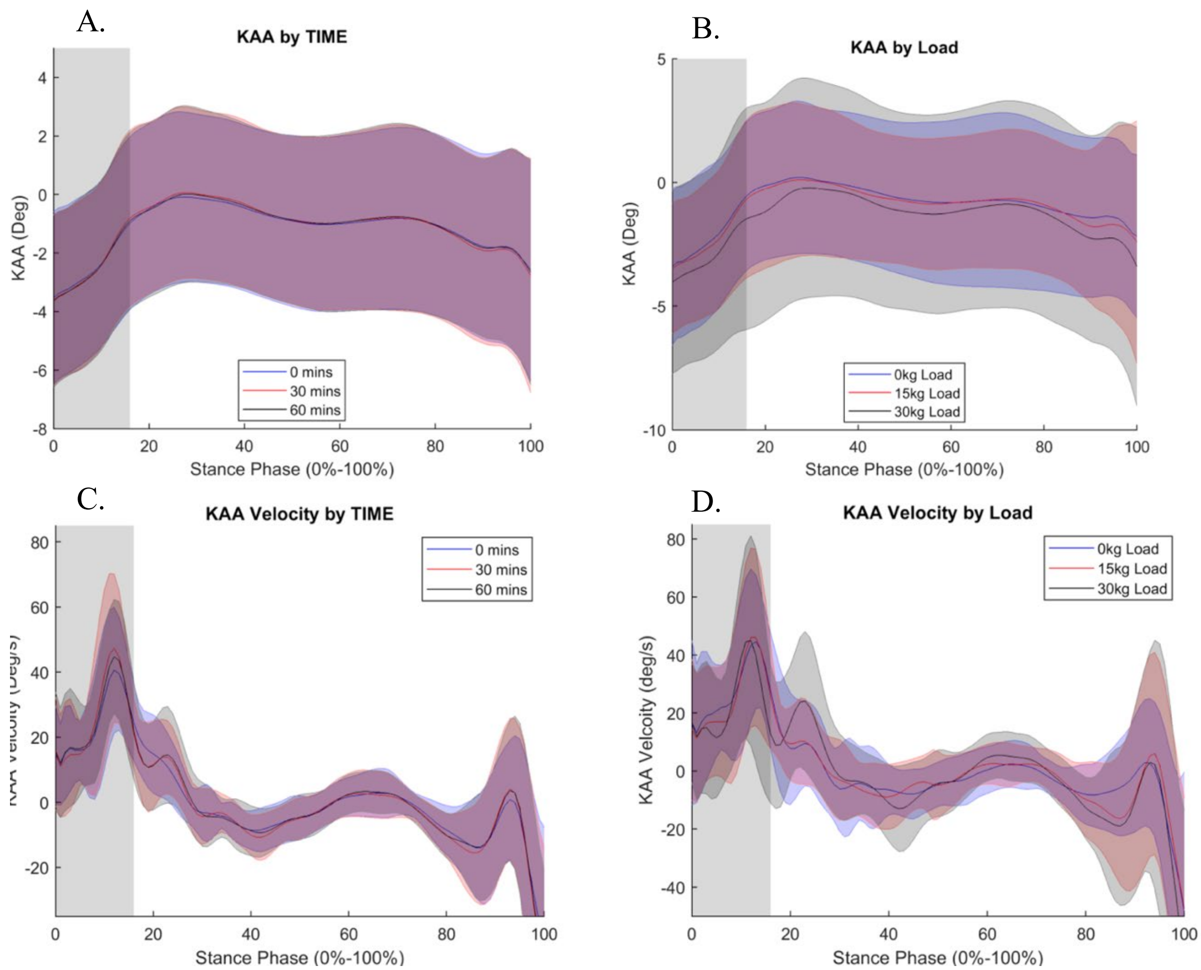


Figure 3.4. Mean ± SD stance phase (0-100%) magnitude of KAA across time (A) and load (B), and velocity of KAA across time (C) and load (D). Grey area depicts first 16% of stance.

<u>Group</u>

The VT participants had greater magnitude and average velocity for both KAA (p=0.003 and p=0.025) and varus thrust (p=0.009 and p=0.007) than the CON. But, no group difference were observed for any KAM measure (p>0.05).

<u>Static Alignment</u>

Static alignment was neither different between the VT and CON group (p=0.412), nor a significant covariate for all knee adduction measures (all: p>0.05)

Discussion

This study sought to examine whether individuals that present varus thrust exhibit greater magnitude and velocity of knee adduction biomechanics during prolonged walking with body borne load. Although the addition of load increased magnitude and velocity of KAM, and walk duration increased magnitude of varus thrust, our hypotheses were only partially supported, as VT participants only exhibited greater magnitude and velocity of knee adduction angle than CON with the lighter 0 kg and 15 kg loads.

The VT participants exhibited larger, faster knee adduction motions than CON participants, which may increase their risk for knee OA development. Specifically, compared to CON, the VT participants exhibited 2.3° and 1.7° greater varus thrust with the 0 kg and 15 kg loads. Varus thrust is reportedly indicative of dynamic knee instability, and may coincide with larger forces transmitted through the joint[91], requiring greater contribution from the knee's passive soft-tissue structures for joint stabilization[39]. The larger varus thrust motion may lead to greater tissue damage at the knee joint, and in fact, individuals that present varus thrust during unloaded walking are four times more likely to develop knee OA[28]. With the light body borne loads, the current VT participants also adopted fasted knee adduction motions than the CON participants. In particular, when walking with the 0 kg and 15 kg loads, VT participants exhibited up to 60% faster average and maximum varus thrust velocity and 40% faster maximum KAA velocity. Considering knee adduction velocity encompasses both speed and direction of the movement, and presented by individuals with radiographically confirmed knee OA[30], the larger and faster knee adduction adopted by VT participants may further elevate their risk for knee OA development. We hypothesize that VT participants may possess a knee morphology, such as greater joint laxity, that pre-disposes them to adopting larger, faster knee adduction biomechanics than CON participants with the lighter loads. Yet, further research is needed to determine if faster knee adduction does, indeed, place more force on the knee's passive soft-tissue structures and elevate knee OA risk when walking with body borne load.

Contrary to our hypothesis, the VT participants did not exhibit larger, faster knee adduction motion with the heavy, 30 kg body borne load than the CON group. In agreement with previous experimental evidence, VT participants decreased magnitude and velocity of knee adduction 46% and 41% with the addition of heavy, 30 kg body borne load; whereas, CON participants increased magnitude and velocity of knee adduction 37% and 33% with the 30 kg load[44]. While the reason only CON participants increased knee adduction motions with heavy body borne load is not immediately evident, we hypothesize it may be related to the neuromuscular control of their knee, or inadequate strength and/or activation of the surrounding knee musculature to prevent increases in knee adduction with the heavy body borne load. Although we hypothesized lower limb alignment would differ between groups, no significant differences in static alignment were currently observed, and therefore, future research may be warranted to focus on neuromuscular control to identify individuals that increase knee biomechanics related to OA with heavy body borne loads.

The addition of body borne load led to larger and faster KAM, but not KAA or varus thrust. In agreement with previous experimental evidence, each incremental addition of body borne load led to a significant increase in magnitude of KAM[92]. Considering KAM is reportedly a correlate for medial knee joint compartment loading, long periods of walking with body borne load may result in the tissue damage that characterizes knee OA[39,93] – particularly considering the additional load may also coincide with faster loading of knee's soft tissues. The current participants also exhibited a significant 11% and 20% increase in maximum velocity of KAM when donning the 15 kg and 30 kg loads during the prolonged walk task. The significant increase in maximum KAM velocity, or rate the external adduction moment was applied to the musculoskeletal system, may require greater muscular effort to stabilize the knee and prevent excessive lateral motion of the joint[11]. Moreover, faster transmission force to the knee joint and associated passive soft-tissue structures may increase risk for tissue damage, as faster loading produces greater deformation of any energy absorption by the tissue[94]. In fact, during unloaded running, faster movements are associated with greater tissue loading and increased risk of lower limb soft-tissue injury[21,95].

Longer walk duration led to larger, but not faster knee adduction motion. Specifically, varus thrust, or lateral knee motion during the first 16% of stance, increased 0.3° after 30 minutes of walking. The physiological demands of long durations of walking[96], particularly with heavy body borne load, reportedly lead to fatigue induced muscular weakness[97,98]. Fatigue induced weakness of the knee's musculature may prevent it from providing active joint stabilization and result in the significant increases in varus thrust currently evident after 30 minutes of walking. Moreover, using greater varus thrust may increase reliance of the knee's passive soft-tissue structures to safely dissipate the impact forces of walking, and elevate the risk for knee injury and OA development. However, considering the current participants exhibited a minimal 0.3° increase in varus thrust towards the end of the prolonged walk task, future research is warranted to determine whether this increase in varus thrust is clinically meaningful and results in greater loading of the knee's passive soft-tissue structures.

Static knee alignment, particularly varus malalignment[25], is purportedly a knee OA risk factor and may increase odds of developing the disease by 2-fold[4,18]. Considering varus malalignment in reportedly related to larger peak KAM[42,43], and larger, faster varus thrust during unloaded walking[30], we hypothesized that individuals with static varus alignment would exhibit greater knee adduction biomechanics when walking with load. Yet, contrary to our hypothesis, static knee varus alignment neither differed between groups, nor exhibited a significant relation to magnitude or velocity of knee adduction biomechanics. Although the current VT participants exhibited a small, insignificant 1.5° difference in static knee alignment compared to CON, the current sample may not be powered appropriately to detect small differences in knee alignment between groups. Future research that tests a larger sample is warranted, as it may be needed to detect differences in static knee alignment between groups and/or to determine whether static alignments impacts knee adduction biomechanics during prolonged load carriage.

This study may also be limited by the current static knee alignment calculation. Currently, static knee alignment was determined using frontal plane knee projection angle. Although using a radiograph may provide less variable knee alignment values than the chosen method, calculating static alignment with the frontal plane projection angle provides alignment values comparable to those quantified using a radiograph[99], and previously exhibited a significant relation with knee biomechanics during dynamic unloaded locomotor tasks[35]. As such, we are confident that the current method of determining static knee alignment was appropriate. Further study of static knee alignment's role in knee adduction biomechanics during prolonged walking with body borne load is warranted.

Conclusion

In conclusion, prolonged load carriage led to increases in magnitude and velocity of knee adduction biomechanics that may elevate risk of knee OA development. The VT group exhibited larger and faster knee adduction motions, and potentially greater OA risk, when walking with the lighter loads (0 and 15 kg); whereas the CON participants exhibited increases in magnitude and velocity of knee adduction when walking with the heavy, 30 kg load not evident for the VT participants. Yet, all participants may increase knee adduction during prolonged walking with body borne load, as the addition of load increased magnitude and velocity of KAM, and walk duration increased magnitude of varus thrust.

CHAPTER FOUR: CONCLUSION

Introduction

This study determined whether body borne load and/or walk duration impact the magnitude and velocity of knee adduction biomechanics, and whether the changes in knee adduction biomechanics differ for individuals with varus thrust and static knee varus alignment. Key findings support the hypotheses that body borne load and walk duration increase magnitude and velocity of knee adduction biomechanics, but only partially support the hypothesis that varus thrust participants will exhibit greater increases of magnitude and velocity of knee adduction biomechanics than control participants.

Key Findings

Participants exhibited larger, faster knee adduction biomechanics during the prolonged load carriage task. Specifically, participants increased magnitude and velocity (maximum) of KAM with the addition of load, and magnitude of varus thrust after 30 minutes of walking. The significant increase in knee adduction biomechanics currently evident during long periods of walking with body borne load may increase an individual's risk of knee OA development, as they are reportedly implicated in the disease pathogenesis at the knee. The varus thrust group exhibited larger, faster knee adduction motions, and potential increase in OA risk, when walking with the lighter (0 kg and 15 kg) loads; whereas, the control participants adopted knee adduction biomechanics related to knee OA, including greater magnitude and velocity of varus thrust, when walking with the heavy (30 kg) loads and after walking for a long period of time.

Significance

This work is the first to document that prolonged walking with body borne load leads to larger, faster knee adduction biomechanics, Specifically, the knowledge that individuals increase magnitude and velocity of knee adduction moment, and magnitude of varus thrust during prolonged load carriage can be used by the military to reduce a service member's risk of musculoskeletal injury and disease development in general, and knee OA specifically. Additionally, this work documented that individuals with varus thrust use larger, faster knee adduction biomechanics and may present greater risk for knee OA development when walking with lighter body borne loads, while individuals without varus thrust may exhibit knee adduction biomechanics related to knee OA when walking with heavy loads for a long period of time. These findings may be implemented by the military to screen for service members that are at higher risk for knee OA, and may lead to substantial reduction in the rates of premature knee OA for service members.

Limitations

The current sample size may be a limitation. Although significant differences in knee adduction biomechanics were observed between groups, the current sample size may not be powered appropriately to detect small differences in static knee alignment between groups. Individuals with varus thrust previously exhibited static knee alignment that was approximately 4 degrees different than healthy controls[30], but in the current study varus thrust and control participants only exhibited an insignificant 1.5° difference in static alignment. Moreover, static knee alignment was not currently observed to be a significant covariate for any knee adduction measure during the prolonged load carriage task, and larger sample size may be necessary to determine whether static alignment impacts knee adduction biomechanics during prolonged load carriage. Additionally, this study may be limited by the current knee alignment calculation. Knee alignment was currently quantified as the frontal plane knee projection angle in accordance to previous literature[90]. Yet, quantifying knee alignment from a radiograph may be less variable than the chosen method and thus, require less participants to detect statistical significance. Quantifying knee alignment with the frontal plane knee projection angle, however, reportedly provides alignment values comparable to a radiograph[99], and previously exhibited a significant relationship with dynamic knee biomechanics during unloaded locomotor tasks[35]. As such, we are confident in our current method of quantifying static knee alignment. Lastly, this study may also be limited by the current participants' load carriage experience. Although the current participants to self-report the ability to carry 75 pounds, they were not required to have prior load carriage experience. Testing participants with load carriage experience may be warranted, particularly during a prolonged load carriage task, as they might exhibit different knee adduction biomechanics than inexperienced participants. However, most service member musculoskeletal injuries occur during basic training when they have limited load carriage experience, and we are currently unaware of any experimental evidence that demonstrated experienced and inexperienced load carriage exhibited different lower limb biomechanics.

Future Work

Prolonged walking with body borne load increased the magnitude and velocity of knee adduction biomechanics. Yet, the specific neuromuscular deficiency that lead to these increases is unknown and future research is warranted to determine the explicit lower limb muscle strength and activation patterns that may mitigate an increase in hazardous knee adduction biomechanics during prolonged walking with body borne load. Considering varus thrust participants exhibited hazardous knee adduction biomechanics with the light loads and control participants with the heavy loads, future research should identify methods for detecting service members at risk for of hazardous knee biomechanics during prolonged walking with both light and heavy body borne load. Although static knee alignment was currently neither different between groups, nor related to magnitude and velocity of knee adduction biomechanics, testing a larger sample size may provide additional insight into service members at risk of knee OA development.

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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

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

Table C.1 Mean ± SD of magnitude and velocity of varus thrust, KAA, and KAM with each of the body borne loads (0 kg, 15 kg, and 30 kg).

	0 kg Load		**15 kg Load**		**30 kg Load**	
	VT	*CON*	*VT*	*CON*	*VT*	*CON*
Varus Thrust Maximum Velocity (deg/s)	65.5 ± 5.4	36.8 ± 5.7	65.1 ± 7.5	39.1 ± 7.9	58.2 ± 7.9	54.3 ± 8.4
Varus Thrust Average Velocity (deg/s) [a]	27.4 ± 1.8	10.9 ± 1.9	25.6 ± 3.0	12.2 ± 3.2	19.8 ± 3.1	18.1 ± 3.2
Varus Thrust (deg) [a]	3.8 ± 0.3	1.5 ± 0.3	3.6 ± 0.5	1.7 ± 0.5	2.7 ± 0.4	2.7 ± 0.5
KAA Maximum Velocity (deg/s) [a]	66.8 ± 6.3	41.1 ± 5.9	68.59 ± 7.9	45.1 ± 7.4	62.1 ± 9.5	57.8 ± 8.9
KAA Average Velocity (deg/s)	27.9 ± 3.2	13.9 ± 3.0	23.0 ± 4.1	12.6 ± 3.8	18.1 ± 2.6	14.6 ± 2.5
KAA Magnitude (deg)	5.1 ± 0.4	2.8 ±0.3	5.2 ± 0.5	3.4 ± 0.5	5.0 ± 0.6	4.2 ± 0.5
KAM Maximum Velocity (Nm/kgm/s) [b]	6.0 ± 0.5	5.8 ± 0.4	7.0 ± 0.6	6.3 ± 0.5	7.5 ± 0.6	7.3 ± 0.6
KAM Average Velocity (Nm/kgm/s)	2.1 ±0.4	2.0 ± 0.4	2.4 ± 0.4	2.0 ± 0.4	2.8 ± 0.5	2.3 ± 0.4
KAM Magnitude (Nm/kgm)[b]	0.4 ± 0.02	0.36 ± 0.02	0.45 ± 0.03	0.42 ±0.03	0.49 ± 0.05	0.47 ± 0.05

[a] Denotes a significant ($p<0.05$) group by load interaction.
[b] Denotes a significant ($p<0.05$) main effect of load.

APPENDIX D

Table D.2 **Mean ± SD of magnitude and velocity of varus thrust, KAA, and KAM at each of the time points (minutes 0, 30, and 60).**

	Minute 0		**Minute 30**		**Minute 60**	
	VT	*CON*	*VT*	*CON*	*VT*	*CON*
Varus Thrust Maximum Velocity (deg/s)	60.7 ± 6.2	40.2 ± 5.8	65.3 ± 8.5	46.0 ± 7.9	62.3 ± 7.2	50.9 ± 6.8
Varus Thrust Average Velocity (deg/s)	23.6 ± 2.1	13.3 ± 1.9	25.4 ± 2.5	14.7 ± 2.4	25.1 ± 2.5	15.4 ± 2.3
Varus Thrust (deg) [a]	3.3 ± 0.3	1.8 ± 0.3	3.6 ± 0.4	2.1 ± 0.4	3.4 ± 0.3	2.2 ± 0.3
KAA Maximum Velocity (deg/s)	63.2 ± 6.4	43.6 ± 6.0	68.9 ± 8.3	47.7 ± 7.9	65.3 ± 7.0	52.6 ± 6.6
KAA Average Velocity (deg/s)	22.2 ± 2.9	12.4 ± 2.8	24.3 ± 2.8	13.4 ± 2.7	22.5 ± 2.6	15.4 ± 2.4
KAA Magnitude (deg)	4.9 ± 0.3	3.3 ± 0.3	5.2 ± 0.4	3.5 ± 0.4	5.1 ± 0.3	3.7 ± 0.3
KAM Maximum Velocity (Nm/kgm/s)	6.6 ± 0.5	6.4 ± 0.5	7.3 ± 0.6	6.4 ± 0.6	6.5 ± 0.4	6.7 ± 0.4
KAM Average Velocity (Nm/kgm/s)	2.4 ± 0.4	2.2 ± 0.4	2.7 ± 0.4	2.0 ± 0.4	2.3 ± 0.3	2.2 ± 0.3
KAM Magnitude (Nm/kgm)	0.43 ± 0.03	0.42 ± 0.03	0.47 ± 0.03	0.41 ± 0.03	0.44 ± 0.03	0.41 ± 0.03

[a] Denotes a significant ($p<0.05$) main effect of time.

APPENDIX E

Static Frontal Plane Knee Projection Angle

Static knee alignment was calculated for the VT and CON participants as the frontal plane knee projection angle in accordance to previous literature[90]. Each groups mean alignment was submitted to a T-test to determine if differences were present between VT and CON.

Results

Static knee alignment for VT was -2.6° ± 3.7 and CON was -4.1° ± 3.4. The 1.5° difference between the VT and CON was not significant ($p=0.412$).

APPENDIX F

Participant Strength and Body Mass

Each participant had hip and knee strength data recorded on an isokinetic dynamometer (HUMAC NORM, CSMI, Stoughton, MA, USA). To record hip strength, each participant performed three maximal isometric hip flexion, extension, and abduction contractions with their dominant limb. For hip flexion and extension contractions, participants stood upright and had their hip secured at 15 degrees of flexion. For hip adduction contraction, participants had their dominant hip abducted to 15 degrees while laying on their non-dominant side[82,83]. To record knee strength, each participant performed three maximal isometric knee flexion and extension contractions with their dominant limb. For each contraction, participants were seated on the dynamometer with their hip and knee secured at 85 degrees and 60 degrees of flexion, respectively[81]. Participants performed three maximal 5 second isometric contractions in each direction (hip and knee flexion and extension, and hip abduction), with 40 seconds of rest between each contraction[84]. The maximum torque produced during the three trials was recorded and normalized to the participant's body mass. Dominant leg was determined by which foot they would kick a ball with[85].

Then, maximal hip and knee strength measures, and body mass, were submitted to analysis to determine if they impacted knee adduction biomechanics recorded during the prolonged load carriage task. Specifically, participant-based means for magnitude and velocity of KAM, KAA, and varus thrust were submitted to a repeated measures ANCOVA to test main and interaction effect of load (*0 kg, 15 kg, and 30 kg*) and time (*minutes 0, 30, and 60*), with maximal hip and knee flexion and extension strength, maximal hip abduction strength, and body mass separately included as covariates. Alpha was set *a priori* $p<0.05$.

Results

Hip and knee strength, but not body mass, were significant covariates for knee adduction (Table F.1). Specifically, when hip abduction strength was accounted for, there was a significant load and time interaction for average KAA velocity ($p=0.008$), as well as a main effect of load on magnitude ($p=0.025$) and average velocity ($p=0.015$) of KAM, and main effect of time on time to peak KAA ($p=0.046$). When knee flexion strength was accounted for, load had a significant effect on maximum KAM ($p=0.014$), KAA ($p=0.019$), and varus thrust ($p=0.007$) velocity, while there was a significant load and time interaction for time to peak KAA ($p=0.003$) and main effect of load for time to peak KAM ($p=0.003$) when knee extension strength was accounted for.

Table F.1 Mean ± SD of maximal knee flexion and extension strength, and body mass.

Maximal Knee Flexion Strength (Nm/kg)	1.24 ± 0.45
Maximal Knee Extension Strength (Nm/kg)	1.69 ± 0.67
Maximal Hip Flexion Strength (Nm/kg)	1.08 ± 0.43
Maximal Hip Extension Strength (Nm/kg)	0.77 ± 0.34
Maximal Hip Abduction Strength (Nm/kg)	0.80 ± 0.37
Body Mass (kg)	71.32 ± 12.15