The Interactive Effects of Load Carriage Magnitude and Gait Velocity on the Tibiofemoral Joint Kinematics in Recruit-Aged Women

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Load carriage is a common military training activity that is important for ensuring warfighters' operational readiness and their ability to safely perform military-relevant tasks. Female soldiers may respond differently to this increased demand and should be included in the existing body of research to inform training and injury prevention practices. **PURPOSE:** To determine the interactive effects of load carriage magnitude and marching velocity on tibiofemoral arthrokinematics in a population of physically active recruit-aged women. **METHODS:** Twelve physically active females walked, ran and force marched on an instrumented treadmill under three loading conditions: unloaded, +25%, and +45% bodyweight (BW). Biplane radiographs were collected of each participant's right knee. Custom model-based tracking software was used to match CT-generated bone models to each pair of synchronized biplane radiographs to recreate *in vivo* bone motion during the dynamic movement. Two-way repeated measures ANOVA were used to compare tibiofemoral kinematics, medial and lateral compartment minimum gap, and normalized compartmental contact path length at 0, 10, 20, and 30% right leg support between load and velocity conditions (α=0.05). Bonferroni-corrected p-values were used to determine the significance of pairwise comparisons between load and velocity conditions (α=0.05). **RESULTS:** Knee flexion increased during forced marching and running compared to walking. Increasing load decreased tibiofemoral gap for the medial and lateral compartments at 10% and 20% of the support phase during running. Joint space decreased by 1.0 mm in both compartments when running as compared to forced marching. A significant interaction between load magnitude and velocity was found for lateral compartment minimum gap, at 30% of stance phase. **CONCLUSION:** Load carriage magnitude was the most influential factor affecting tibiofemoral joint space, while increased gait velocity affected knee flexion and dynamic joint space at select portions of the early support phase.

Limited interaction effects were observed between the increase in load carriage and locomotion velocity.

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1.0 Introduction

With the United States military opening combat roles to female warfighters, special care must be taken to ensure that the nations' fighting force can continue to train safely and effectively while maintaining combat readiness. The opening of these positions to women indicates the need for extensive, in-depth research into the demands placed on recruits during physical training activities in order to minimize injury and optimize performance. Areas of particular concern are training activities which incorporate heavy load carriage and any resulting adaptations in gait kinetics and kinematics during these activities. Warfighters in combat roles are required to carry heavy loads during long-distance training exercises, sometimes over 100 lb.(Attwells, Birrell, Hooper, & Mansfield, 2006; Wang, Frame, Ozimek, Leib, & Dugan, 2012) Given that the average woman has a lower body weight than the average male, a fixed load of equipment could be proportionally larger for female soldiers and potentially have a greater effect on their biomechanics and stability. Additionally, marching velocity plays an important role in military operations, as soldiers are often encouraged to complete marches at a pace greater than the pace at which they would normally transition from walking to running. This "forced march" allows soldiers to cover ground quickly but the effects of loading on the gait parameters of this high-speed walking are understudied and are important to consider in order to prevent injury in military populations. This long-distance running, coupled with walking and forced marching while carrying additional loads are an integral part of military training programs and can often amount to 200 miles of cumulative distance travelled over the course of a single week.(Wang et al., 2012) While military populations already show a high incidence of lower extremity overuse injuries during training, female warfighters are at a greater risk of developing lower extremity stress fractures due to factors such as anatomical differences, disparities in physical fitness at the onset of BCT, and biomechanical response to load carriage.(Bell, Mangione, Hemenway, Amoroso, & Jones, 2000; Bullock, Jones, Gilchrist, & Marshall, 2010; Grier, Canham-Chervak, Anderson, Bushman, & Jones, 2017) As a consequence, it is important to assess lower extremity biomechanics in a female recruit-age population to ensure that they can safely and effectively meet the military's operational standards. Since the knee joint is the site of approximately 50% of noncombat-related injuries, focusing on the interaction between femoral and tibial articulating surfaces, or arthrokinematics, during high-impact activities such as marching while carrying a load could provide valuable information to military personnel regarding preventing injuries and developing effective training methods for all warfighters.(Seay, 2015)

1.1 Military Load Carriage

1.1.1 Training Activities

An important aspect of military training is allowing the warfighters to acclimate to the amount of equipment and the magnitude of load they will be expected to carry in day-to-day operations. These loads can vary depending on the training exercise or environment, as warfighters generally carry a large amount of gear and supplies and remove extra weight when coming in contact with hostile forces.(J. J. Knapik, Reynolds, & Harman, 2004) At maximum, warfighters are expected to be able to carry up to 70% of their bodyweight.(Wang et al., 2012) However, in recent years the military has been altering procedures to allow soldiers to carry as little equipment and remain as mobile as possible while still maintaining warfighter health, safety, and operational performance.(J. J. Knapik et al., 2004) Reducing load is not always an option, but technological developments in the form of waist packs, internal and external frame backpacks, hand carts, and lighter gear have helped alleviate some of the burden. Physical training with a pack allows warfighters to develop greater mobility and improve their overall strength and aerobic fitness. Analyzing the effects of carried loads on the gait biomechanics of female soldiers will provide additional information regarding the warfighter's capability to maintain mobility at various speeds and loading conditions.

1.1.2 Gait Biomechanics

Previous research has shown that simply running in combat boots puts soldiers at an increased risk for tibial stress fracture injuries, ankle inversion injury, Achilles tendon strain, and shin splints, among other injuries of the lower extremity.(Dixon, Waterworth, Smith, & House, 2003; Nicola & Jewison, 2012) However, there is a lack of research regarding how additional loading and forced marching velocities affect knee kinetics and kinematics. With the increased range of motion required during running, it is possible that the addition of load carriage could significantly increase stresses at the knee joint and result in damage or an increased risk of injury. At a set velocity, increasing the load carried has been shown to result in a decrease in stride length, an increase in percentage of double support time, an increase in time spent in swing phase, and an increase in step frequency.(Birrell & Haslam, 2009; Seay, 2015) Although these shorter, faster steps allow the load carriers to maintain their walking velocity, it may be more biomechanically optimal to transition to a running gait once a certain velocity is reached. Additionally, considering the dynamic joint space and cartilage contact areas during running, walking, and loaded marching tasks are essential to fully understand the mechanisms of long-term knee joint degeneration.

1.1.3 Considerations for Female Warfighters

Additional considerations for female warfighters are necessary due to size, weight, and anatomical discrepancies between men and women. In general, a larger warfighter may be able to carry more load than a smaller warfighter due to a larger amount of muscle and bone mass.(J. J. Knapik et al., 2004) Since the average woman has a lower body mass, it is possible that this could negatively impact load carriage capabilities in an untrained female population. Additionally, increased loads may affect gait kinematics in women differently than they do males. Women generally walk with a shorter stride length and greater stride frequency compared to men because, on average, their legs are shorter and pelvic girdle is wider, leading to more significant kinematic changes with the introduction of a load, such as a weight vest or backpack.(Kelly, Jonson, Cohen, & Shaffer, 2000; J. J. Knapik et al., 2004) A study by Knapik et al found that increased load further decreases stride length in females with a corresponding increase in stride frequency.(J. Knapik, Harman, & Reynolds, 1996) In contrast, male subjects were able to endure the additional load with no significant differences in stride frequency. Additionally, double support time during the walking gait cycle was increased in the loaded condition for female subjects.(J. J. Knapik et al., 2004)

As indicated by previous research, load carriage training will potentially have a more significant effect on the biomechanics of female recruits. However, no studies have been conducted in this specific population to understand how velocity and load affects tibiofemoral kinematics. Analyzing tibiofemoral translations and rotations, contact path and areas, and joint space during loaded locomotion could provide insight into potential injury mechanisms at the knee that are specific to female warfighters and inform training and operational strategies for this population.

1.1.4 Injury Risk

Previous research has shown that carrying a heavier load leads to increased risk of musculoskeletal injury.[6] In a military population, individuals are required to carry these loads over distances of up to 200 miles per week in training, compounding the damage done by an increased impact force to the lower extremity.(Wang et al., 2012) The knee joint is a common injury site during military physical training. In a study documenting running-related injuries in young male recruits, Grier et al found that 31% of running-related injuries were located at the knee, followed by the ankle (20%), lower leg (11%), foot (7%), and lower back (6%).(Grier et al., 2017)   With the addition of a backpack, it is possible that the incidence of injury at the knee will increase in comparison to running without a backpack due to differences in tibiofemoral kinematics under a loaded condition and an increase in vertical ground reaction forces.(Simpson, Munro, & Steele, 2012) Han et al. found that increased load resulted in an increased external flexion moment at the knee.[7] An increase in load on the knee joint structure has the potential to lead to pain and injury if the individual is not sufficiently conditioned. Jones et al investigated changes in knee abduction and flexion moments at the knee joint in individuals walking at a self-selected pace with no pack, 12 kg pack, and 24 kg pack.(Jones, Bovee, Harris, & Cowan, 1993) In all conditions, peak knee abduction moment and peak flexion moment increased with the increase in load.[6]

1.2 Definition of the Problem

The introduction of women into combat roles poses many unique challenges in order to safely accommodate female warfighters during training an d operational activities and ensure military doctrine is informed of any potentially injurious training practices for both genders in terms of military training activities. Since all recruits are held to the same standard, regardless of sex or size, it is important to consider how female warfighters respond to situations of intense physical activity common to military training. Specifically, recruits are subjected to loading conditions up to 70% of their bodyweight and required to march at a high velocity for extended periods of time.(J. Knapik et al., 1996) Such cumulative loading during training activities can lead to an increased risk of lower extremity overuse injuries, to which previous research has shown women are already more susceptible.(J. J. Knapik et al., 2004) Previous studies have investigated the effect of load carriage on walking and running gait kinematics, ground reaction forces, and other variables, but additional research is needed to comprehensively address the effect of load carriage and gait velocity on tibiofemoral arthrokinematics in women.

1.3 Purpose of the Study

The purpose of this study is to investigate the interactive effects of load magnitude and gait velocity, determining how these conditions affect tibiofemoral arthrokinematics in healthy recruit-aged females. A secondary aim of this study is to determine if load carriage magnitude affects these biomechanical variables, and show any differences between walking, running, and forced march conditions.

1.4 Specific Aims and Hypotheses

Specific Aim 1: To determine the effects of increased load carriage on tibiofemoral arthrokinematics (bodyweight (BW), +25%BW, +45%BW).

Hypothesis 1: Increased load carriage will increase flexion, decrease tibiofemoral dynamic joint space, and increase subchondral contact path length.

Specific Aim 2: To determine the effects of different locomotion velocities on tibiofemoral arthrokinematics (walking, running, and forced march at a high velocity).

Hypothesis 2: Increased velocity will increase flexion. The forced march condition will elicit the greatest kinematic changes at the knee joint. The tibiofemoral contact path will shift posteriorly with increased locomotion velocity.

Specific Aim 3: To identify the interactive effects of load carriage and marching velocity on knee arthrokinematics.

Hypothesis 3: There will be a significant interaction of load carriage and locomotion velocity for flexion, joint space, and contact path.

1.5 Study Significance

Although previous studies have addressed the biomechanical consequences of walking and running during a loaded condition, there is a lack of research specific to a female population and focused on the dynamic interaction between the femur and tibia. This study will describe important components of knee joint biomechanics in a female recruit-aged population during load carriage at different velocities. The primary aim of this study is to investigate the tibiofemoral joint kinematics and stresses elicited by different loading conditions, which could indicate an optimal training load that allows warfighters to remain mobile and move through their environment in a safe manner without injury. The secondary aim of this study is focused on the relationship between tibiofemoral joint kinematics and marching velocity. This could indicate if it is more biomechanically optimal for warfighters to perform their ruck marches at a specific velocity. Overall, this study could potentially provide information to help reduce the incidence of lower extremity injuries and potential long-term joint degeneration in a female recruit-aged population. It may also provide insight into potential training methods that will help soldiers compensate for any biomechanical limitations intrinsic to standard combat footwear, marching velocity, or load carriage conditions.

2.0 Review of Literature

2.1 Military Load Carriage

Warfighters are required to utilize different load carriage systems to transport equipment and supplies during training and operations. Proper design of these systems should maximize comfort and efficiency and mitigate any localized stresses that could potentially lead to injury, while recommendations from military leaders for safe training and operational loads should be informed by research regarding load carriage injuries and risk factors. Additionally, the introduction of women into combat roles necessitates increased research efforts into female-specific responses to highly demanding tasks, such as sustained load carriage. Doing so may help inform military doctrine and mitigate any gender-injury disparities that exist within military populations.

2.1.1 Load Carriage History and Training

Traditionally soldiers have been expected to remain mobile while carrying heavy loads, requiring both strength and muscular endurance.(Nindl, 2015) Up until the 18th century, troops carried light loads of essential combat items, usually less than 15 kg, with extra equipment moved by auxiliary methods of transport such as horses, hand carts, or assistants.(Seay, 2015) This practice became less common in the following years with individual soldiers becoming responsible for carrying more of their own equipment, rations, and weapons. While modern day militaries have the technological ability to lighten soldier burden through effective auxiliary transport and engineering more efficient combat materials, today's soldiers still carry increasingly heavy loads. The average U.S. soldier in a combat role carries a load averaging around 45.5 kg (100 lb) but was even greater in certain combat units.(Schuh-Renner et al., 2017; Seay, 2015) Although the U.S. military has attempted to implement a method of "load echeloning" in an effort to alleviate soldier burden, where carried loads are divided into categories based on the amount of equipment and mobility necessary for the intended operation, the actual loads carried by warfighters significantly exceed the recommended magnitude.(Nindl, 2015) The lightest carried load, or the fighting load, often accounts for around 35% of soldier body mass, while marching loads such as the approach march load and emergency march loads account for 57% and 78% of an individual's body mass, respectively.

"Road" or "ruck" marching is a common endurance training activity that can be performed on any type of terrain with the goal of moving a specific distance while carrying a load in a rucksack or backpack.(Schuh-Renner et al., 2017) This exercise aids in a soldier's physical conditioning and load carriage acclimation as well as enhances readiness for operational activities.(J. J. Knapik et al., 2004) Loads carried during these training sessions can include all the necessary equipment, ammunition, weapons, and survival tools necessary for combat situations.(J. J. Knapik et al., 2004; Schuh-Renner et al., 2017; Seay, 2015) A forced-cadence march is commonly used to reach areas less than three miles away, which may prove problematic for smaller soldiers attempting to maintain the same step length and rate as their taller counterparts.(Seay, 2015) Overstriding puts an individual at increased risk for lower extremity stress fracture and could disproportionally affect female warfighters due to their on average smaller size.(Kelly et al., 2000) Training loads for ruck marches can range from 40 lb to over 100 lb and can last for distances ranging from two to twelve miles.(Schuh-Renner et al., 2017)

March velocity can vary based on the terrain being traversed but a common task is to complete a twelve mile march in three hours (4 mph).(Schuh-Renner et al., 2017)

2.1.2 Military Load Carriage Recommendations

As loads carried by individual warfighters have increased, so have injury rates in the U.S. Army.(Seay, 2015) Knee and ankle injuries are the most common overuse injuries in military personnel and tend to be not related to combat activities.(Seay, 2015; Sell et al., 2010) Chronic lower extremity injuries result in a loss of duty time and a high cost in terms of medical resources, so it is important to develop load carriage practices that mitigate injurious forces and help maintain soldier mobility. The Army Public Health Center's Injury Prevention Program provides a set of recommendations regarding the prevention of road marching-related injuries. Important considerations for reducing these injuries include proper use of equipment, utilizing a progressive training program, and maintaining a healthy lifestyle.

When carrying heavy loads, energy expenditure increases with an increase in load magnitude, so it is essential to find an optimal load distribution that does not induce additional energy cost.(Seay, 2015) A properly fit backpack and footwear system is essential for a recruit beginning their road marching training. The U.S. Army recommends a lightweight external frame pack system with the use of straps and belts if needed to help redistribute loads.(12-008-0416., 2016) A properly adjusted hip strap can account for 30% alleviation of the carried load and should be used if the straps themselves do not put significant pressure onto the hips and cause further injury.(12-008-0416., 2016) Load location and distribution should also be considered to allow for the most natural movement of the soldier during the march. In general, it is most energy efficient to carry loads as close as possible to the body's center of mass.(J. J.

Knapik et al., 2004; Seay, 2015) Common load carriage systems, such as the backpack or the double pack, can also be loaded in such a way that reduces the energy cost to the individual. Seay et al. found that carrying a heavy load in a backpack high up on the individual's torso and close to their body was +25%BW more energy efficient than carrying the same load low in the backpack and away from the carrier.(Seay, 2015) Use of double packs, which have a backpack component and an anterior component which distributes almost half of the load in front of the torso, can also be useful in helping minimize trunk lean and deviations from normal walking patterns.(J. J. Knapik et al., 2004) However, these load carriage systems may not be operationally sound in some scenarios, as they can limit soldier mobility and field of vision. A combination of load carriage systems such as a backpack with a trunk vest and hip belt can help distribute the load more evenly, reduce energy cost, and improve body posture without the restrictions of the double pack.

A progressive training program should be utilized when introducing recruits to road marching activities to allow them to build up physical fitness, avoid overuse injuries, and eventually progress to carrying operational or mission-specific loads. Increasing load magnitude, march speed, march distance, and frequency of march training too rapidly should be avoided. The training program should also consider the terrain of the march, and adjustments to speed, distance, and load carried should be made for particularly difficult terrain such as hills or snow, as they may increase injury risk.(12-008-0416., 2016) Adjustments to the load magnitude can also be made to allow the soldiers to carry only mission essential items. This reduced load can help recruits to travel faster over a greater distance without carrying unnecessary equipment.(12-008-0416., 2016) March speeds greater than 4 mph are not recommended since running with a carried load has the potential to increase the soldier's risk of sustaining an injury, particularly to the lower extremity.(12-008-0416., 2016) A 1996 study of Navy recruits undergoing basic training showed that female soldiers who served as road guards during marches and were required to run short intervals for the duration of the march had a higher risk of sustaining a stress fracture.(Kelly et al., 2000) Rotating the road guard position between marches and discouraging these short spurts of running, especially while loaded, could help lower the incidence of stress fractures associated with this type of march.

Lastly, improving overall muscular strength, aerobic fitness, and endurance may decrease the risk of sustaining a road marching injury. A soldier's regimented physical training program should be well-balanced exercises to achieve fitness in each of these areas, as well as incorporating agility and balance training. Avoiding excessive running and marching distances during training activities is recommended to avoid overuse injuries. Overall, in the effort to make military load carriage more physiologically and operationally sound, considerations should be made for improving load carriage training programs, advocating for the use of auxiliary mechanisms of transporting equipment, and decreasing the load carried by individual soldiers to maintain a healthy and effective fighting force.(Nindl, 2015)

2.2 Musculoskeletal Injury in the Military

Musculoskeletal injury (MSI) is extremely common among military populations due to the rigorous nature of training and performing operational tasks.(Sell et al., 2010; Wang, Frame, Ozimek, Leib, & Dugan, 2013) Even minor injuries can affect warfighters' operational readiness. Common lower extremity injuries related to military training are foot blisters, low back injuries, metatarsalgia, upper brachial plexus palsy, knee pain, and lower extremity stress fractures.(J. J. Knapik et al., 2004) Demonstrated risk for stress fractures in military populations include female sex, Caucasian ethnicity, stature, prior inactivity, load carriage distance, and walking style, among others.(J. J. Knapik et al., 2004) Running, in particular, correlates with lower extremity musculoskeletal injuries during military training, so it is important to analyze how running and load carriage activities may interact and affect the incidence of injury in military populations.(Bullock et al., 2010; Sell et al., 2010; Smith & Cashman, 2002)

2.2.1 Military Musculoskeletal Injury Epidemiology

It is especially important to consider the quantity and types of injuries occurring in military populations in order to allocate healthcare resources, promote training strategies and practices that allow for injury prevention, and maximize operational readiness. Historically, an increase in loads carried by U.S. soldiers has been associated with an increase in MSIs, with the ankle and knee being the most commonly injured joints.(Seay, 2015; Sell et al., 2010; Smith & Cashman, 2002) Injury to the knee is the most common of all noncombat MSI and accounts for approximately 50% of all noncombat injuries, but soldiers in a combat arms Military Occupational Specialty (MOS) are at a much higher risk of sustaining an overuse knee injury than any other MOS.(Seay, 2015)

Injuries during BCT are common due to the abrupt increase in amount and frequency of physical activity and the regimented, strenuous training protocols that are employed.(Bullock et al., 2010) Unfamiliar tasks, such as formation marching, running, calisthenics, and military drill, can also contribute to the increase in injury incidence during this period. However, this also makes BCT an ideal environment for studying training-related injuries since soldiers live and work in identical conditions and perform the same regimented conditioning programs, allowing for equal exposure to risks. Bell et al. identified that among trainees followed during BCT, women were at a greater risk of sustaining a training-related injury than their male counterparts when controlling for exposure.(Bell et al., 2000) The risk of sustaining a serious time-loss injury (at least one day of lost duty) was 2.5 times greater in female trainees than in male trainees.(Bell et al., 2000) At the initiation of BCT the male soldiers had significantly higher entry-level measures of physical fitness than women, except for flexibility. The relatively decreased fitness level of women compared to men may contribute to the increased rate of injury in female trainees, who experienced twice as many injuries as men throughout the study period.(Bell et al., 2000) This fitness gap was narrowed as the result of BCT and fitness-adjusted injury rates show no significant gender differences, with women showing improvements to their physical fitness at a rate twice that of men.(Bell et al., 2000) Therefore the gender-injury relationship likely explained by fitness level, rather than gender differences, and injury risk may be reduced through modified training programs.

As expected, many studies focus on the incidence of MSI, which make up a large proportion of time-loss injuries. In a study of 96 female Navy recruits, 181 stress fractures were identified during the nine-week basic training period, for an injury rate of 19.71 injuries per 1,000 recruits. The stress fractures occurred largely in the lower extremity with injuries to the tibia (53%), metatarsals (21%), pubic rami (14%), calcaneus (7%), and femur (5%).(Kelly et al., 2000) Another randomized, retrospective medical record review showed the annual incidence of injury was 95 injuries per 100 soldiers per year with 372 injuries representing 56% of sick call diagnoses. Physical training was linked to 50% of these injuries, of which 30% were linked to running specifically.(Smith & Cashman, 2002) These results are supported by other studies, which cite that out of 412 injuries reported from the studied infantry battalions running in PT was responsible for 27% of injuries (n=113 out of 412) with 23% of injuries to the knee, 26% to the back, and 18% to the ankle.(Schuh-Renner et al., 2017; Sell et al., 2010)

Although their role in sustaining effective combat operations is essential, the addition of road marching activities into a soldier's physical training regimen may also contribute to the number of injuries observed in military trainee populations. A study by Schuh-Renner et al. identified that road marching was associated with 23% of injuries reported by two infantry battalions (n=96 of 412 reported injuries) and that half of the injuries resulted in temporary duty restrictions. Soldiers in these battalions marched on average five times per month and about 7.4 miles per session, carrying an average load of 44 lbs (20 kg).(Schuh-Renner et al., 2017) Injuries per mile for road marching in this population was 5.9 injuries per 10,000 miles marched, while injuries per mile for running along was significantly lower at 3.3 injuries out 10,000 miles run.(Schuh-Renner et al., 2017)

2.2.2 Injury Types

Two main types of injury which occur during military marching or running activities are overuse injuries and gait-biomechanics related injuries. Overuse injuries result from repeated loading followed by inadequate recovery. Following repetitive stresses such as running or road marching the bone tissue remodeling process begins, but is temporarily weakened prior to new bone formation and leaves the area more susceptible to injury.(J. J. Knapik et al., 2004) In a military population, overuse injuries can occur from either training or performing operational tasks. During Basic Combat Training (BCT), the trainee is subjected to a major increase in workload and physical activity which may be unfamiliar.(Seay, 2015) Activities performed during BCT aim to bring the trainees to a base level of physical fitness and include an increase in running frequency and carrying operationally relevant loads during ruck marches. However, overuse injuries are not limited to an untrained population. Although active duty soldiers are highly trained and physically fit, repetitive activity and load carriage can still increase the soldier's risk of sustaining an injury.

Injuries resulting from gait biomechanics should also be considered when analyzing military populations. In the general population, overstriding is considered a risk factor for lower extremity stress injuries. In a military population, injuries resulting from overstriding are of great concern since the warfighters spend much of their time on their feet or marching. In forced-cadence road marching the march pace for the entire group is set by the drill sergeant and the group maintains the pace for the duration of their march. This tactic is a commonly used training exercise to reach an objective less than three miles away(Seay, 2015). If the set cadence is more optimal for a taller trainee with a longer stride, a smaller trainee with a shorter stride may struggle to maintain cadence and consequently begin overstriding. In a study of 13 civilians walking with a 20 kg load at a self-selected pace, a slower pace, and a higher pace, Seay et al. found that an increase in step length was correlated to an increase in anteroposterior ground reaction forces and concluded that overstriding resulted in a 20% increase in stress at the knee extensor musculature.(Seay, 2015) This puts shorter recruits who need to take longer steps to keep pace at an increased risk for musculoskeletal injury and may disproportionately affect female recruits due to biological size differences. Decreasing stride length while marching or having shorter recruits set the pace may reduce incidence of stress fracture in these populations.(Kelly et al., 2000) An additional risk factor was identified in a 1995 study of female Navy recruits undergoing nine weeks of military and physical training. While investigating the gender disparity of pelvic stress fractures in military recruits Kelly et al. found that individuals who often performed road guard duty were at a greater risk of stress fracture than other recruits of a similar fitness level.(Kelly et al., 2000) During formation marches recruits serving as road guards were required to run ahead of the group to halt traffic at an approaching intersection and then promptly return to their position around the perimeter of the battalion. Overall, recruits who sustained a pelvic stress fracture during the study period were more likely to be shorter, lighter, marching in the rear of their battalion, serving as road guards, or feeling that they were overstriding during walking, running, or marching. Following this study, a policy of prohibiting road guards from running and rotating the road guard position daily was implemented, resulting in a decrease in the overall stress fracture rate in female recruits (3.5% to 1.3%) and an 80% decrease in the rate of pelvic stress fractures.(Kelly et al., 2000)

2.2.3 Burden of Injury

Injury and illness not only affect the operational readiness of a unit but also incurs significant costs in terms of medical care and days of limited duty.(Wang et al., 2013) Several studies have shown the substantial training hours lost to musculoskeletal injuries. According to the Armed Forces Injury Prevention Work Group injuries in the U.S. Army total a loss of over 550,000 workdays per year. Of these injuries, more than 75% are caused by physical training.(Smith & Cashman, 2002)

In a randomized retrospective medical review, an annual injury incidence of 95 injuries per 100 soldiers per year was identified, accounting for 56% of sick call diagnoses during the study period.(Smith & Cashman, 2002) 50% of these injuries were linked to physical training exercise and 30% of the PT-associated injuries were related to running.(Smith & Cashman, 2002) A study by Smith et. al in 1998 showed that injuries caused nearly 10 times the number of limited duty days as illness, with 60% of injured recruits unable to immediately return to full duty following their injury.(Smith & Cashman, 2002) Of the injuries that occurred during the study period, injuries to the lower extremity that were associated with running caused the injured individual to spend seven times more days on altered PT than those with non-running injuries.(Smith & Cashman, 2002)

The resulting period of limited duty as soldiers complete rehabilitation programs can be extremely costly to the military.(Sell et al., 2010) Rehabilitation time for a more serious injury such as a lower extremity stress fracture can range from 4 to 8 weeks, but it is important to complete the protocol in full to maintain the soldier's health and future military opportunities.(Kelly et al., 2000) According to the Department of Veterans Affairs orthopedic conditions account for around 53% of all disabilities incurred by the army.(Smith & Cashman, 2002) In 1994 the Department paid $346 million per month in medical disability for orthopedic injuries across all military services.(Smith & Cashman, 2002) In addition to these disability payments additional training costs are incurred to replace medically discharged personnel.(Wang et al., 2013) Besides saving money, reducing the incidence of injury in training and operational scenarios could increase morale among trainees and servicemembers.

2.3 Gait Biomechanics

2.3.1 Gait Cycle

When considering the effects load carriage and velocity on knee biomechanics, it is important to consider the basic biomechanical concepts of the walking and running gait cycle. A person moves through two distinct stages in the walking gait cycle: a stance phase where the foot contacts the surface and a swing phase where one foot is planted and the opposite foot swings forward to complete a stride.(Brown, O'Donovan, Hasselquist, Corner, & Schiffman, 2014; Nicola & Jewison, 2012; Seay, 2015) In this manner, either one or both extremities will be in contact with the ground at any given time. While walking a single limb will be in stance phase for approximately 60% of the gait cycle. The running gait cycle differs from the walking gait cycle due to the addition of a float phase where both feet leave the ground, due to overlapping swing phases.(Nicola & Jewison, 2012; Seay, 2015) The float phase occurs twice during the gait cycle leading to a lower percentage of stance time (approximately 50% of the gait cycle is spent with a limb in stance phase) when compared to a typical walking gait pattern.(Nicola & Jewison, 2012)

Understanding the components of the gait cycle is critical for accurate analysis of kinematics. During the mid-stance phase of running increased knee flexion helps maintain stability by lowering the body's center of mass.(Birrell & Haslam, 2009) No net mechanical work is performed during steady-state running (running at a constant speed) with the stance limb alternating between energy generation at push off and energy absorption at foot strike and load acceptance.(Brown et al., 2014) During mid-stance the extended limb helps maintain stability by lowering increasing knee flexion to lower the body's center of mass.(Birrell & Haslam, 2009; Brown et al., 2014) By analyzing knee joint kinematics during loaded marching and running activities it is possible to identify factors which contribute to increasing stability at the joint during these extreme conditions.

2.3.2 Gait Transition

The transition from running to walking is notable due to its implications for neuromuscular control of the lower extremity and for carrier stability and neuromuscular control of lower extremity. This transition is realized in the final step immediately prior to the first occurrence of flight phase as the gait pattern shifts from walking's double-support stance phase to running's signature flight phase.(Segers, Lenoir, Aerts, & De Clercq, 2007) During this transitional step the stance leg exhibits greater flexion at the knee while the ankle and hip provide the extra mechanical energy (approximately three times as the amount used while walking) necessary to propel the body into the flight phase of running.(Segers et al., 2007) Additionally, when carrying a heavy load the amount of energy required to transition from a walking to a running gait pattern may be required.(Brown et al., 2014) The speed at which an individual initiates the transition is dependent on factors such as body size, training, and other parameters. In a study of thirteen physically active females performing cyclic walk-to-run (WRT) and run-to-walk (RWT) transitions, Segers et al. determined that transition speeds were 2.16 ± 0.12 m*s^{-1} for WRT and 2.19 ± 0.12 m*s^{-1} for RWT.(Segers et al., 2007)

2.3.3 Effect of Loading

The addition of a body-borne load requires the carrier to adapt their biomechanics in order to maintain stability while walking or running. These changes manifest as spatiotemporal changes to the carrier's running or walking gait cycle and kinematic changes in the torso and lower extremity. At a set speed, increasing the load carried results in a decrease in stride length, an increase in percentage of double support time, an increase in time spent in swing phase, and an increase in step frequency.(Birrell & Haslam, 2009; Seay, 2015) Although these shorter faster steps allow the load carriers to maintain their walking velocity, it may be more biomechanically optimal to transition to a running gait once a certain velocity is reached.

Another important factor to consider when analyzing the biomechanical effects of loading is and load distribution. Studies utilizing a weighted vest with equal load distribution anteriorly and posteriorly can help isolate any biomechanical changes that are due to the load magnitude itself, rather than its placement. Studies attempting to mimic an operational military load carriage scenario often incorporate a standard pack-borne load and a weighted simulated assault rifle held in front of the body during motion testing. This addition may increase the study's validity in terms of its ability to relate their results to military readiness, but this anterior load alters lower extremity kinematics and can confound effects of the backpack-borne loading condition. Loads carried around the torso in a backpack or other load carriage mechanism results in an increase in forward trunk lean as the carrier's center of mass sits posteriorly and can occur with loads as light as 6 kg.(Attwells et al., 2006; Seay, 2015) For heavier loads (>40 kg), such as those carried by soldiers in combat roles, this lean can be as extreme as 11˚ relative to the horizontal plane.(Seay, 2015) In a study comparing the kinematic effects of different load carriage systems, Attwells et al. found significant differences in trunk angle during the stance phase of gait in response to increasing loads using a backpack and while holding a light antitank weapon.(Attwells et al., 2006) This compensation aids in stability and helps keep the load centered over the carrier's base of support.(Seay, 2015)

Additional changes as a result of increased loading occur further down the kinetic chain. Maximum hip flexion increases with an increase in load to compensate for the increased trunk lean, but changes in knee flexion angle and range of motion are not as universally observed and can vary widely between studies.(Seay, 2015) Observable differences between studies may be due to the load distribution and equipment type used, as well as the subject's physical activity level and previous load carriage experience. One study found that sagittal plane knee range of motion (ROM) significantly increased with additional loading due to increased flexion at heel strike and greater extension at toe-off.(Attwells et al., 2006) This increased flexion during load acceptance following heel strike may be a protective mechanism to aid in absorbing impact.

2.3.4 The Knee Joint Under Load

While the spatiotemporal effects of running while loaded have been widely studied, there is less available research regarding lower extremity 3D kinematics and even fewer studies that explore these effects on female subjects. Although findings vary between studies due to differences in load carriage systems, gait speeds, and the subject's load carriage experiences, it is generally agreed that the knee is kinematically sensitive to increasing or decreasing loads during the weight acceptance portion of early stance phase.(Seay, 2015) In a 2009 study of males with significant load carriage experience, Birrell et al. found significant decreases in sagittal plane knee flexion and extension with increased load during walking (target speed 1.5 m/s, ±5%).(Birrell & Haslam, 2009) A similar study analyzing preferred step rate found that between no additional load and a load of 55 kg joint moments at the knee increased 118%.(Seay, 2015) Similarly, Jones et al investigated changes in knee abduction and flexion moments at the knee joint in individuals walking at a self-selected pace with no pack, 12 kg pack, and 24 kg pack. In all conditions, peak knee abduction moment and peak flexion moment increased with the increase in load.[6] This increased stress around the tissues of the knee joint has the potential to induce injurious forces and should be further studied to quantify the relationship between gait speed and loading magnitude with the biomechanics of the knee joint.

Loading of the knee joint is realized in each step of the gait cycle, whether the individual is carrying an additional load or simply their own body weight. As the knee joint is loaded the cartilage contact area may change in size or location based on factors such as the degree of flexion of the knee, the amount of ground reaction force, the terrain, or the gait speed. In a study of 44 healthy subjects walking on a treadmill it was determined that medial compartment cartilage contact area was greater than the lateral compartment during level walking and downhill running, as measured using a biplane radiography system.(Akpinar et al., 2019) The medial compartment is also the location of the peak contact stresses during simulated gait.(Gilbert et al., 2014) Contact path can also be affected based on gait speed, with longer, more posterior tibiofemoral contact paths during downhill running versus walking.(Akpinar et al., 2019) Understanding the knee's arthrokinematics when influenced by loading and velocity changes is essential to understanding potential injury mechanisms at the knee during military activities.

2.3.5 Measurement of Knee Kinematics

Quantifying knee kinematics during dynamic movements is important for identifying potentially pathological or injurious biomechanics. For example, patients with ligament deficiencies are symptomatic under dynamic conditions, so static analysis may not accurately reflect functional disabilities.(Ino et al., 2015) Similarly, static analysis of the knee joint under load may miss important translation and rotational differences that occur during different portions of the gait cycle. Methods for measuring dynamic joint motion include skin-based optical or inertial motion capture systems, intracortical pins or surgically implanted tantalum beads, robotic simulations using cadaver specimens, and biplane radiography. While cadaver-based simulations are useful for investigating loading during static positions it is less useful when attempting to quantify dynamic motions, since the simulations do not accurately represent contributing muscle forces, loading conditions, or natural motion of the segment.(Bey, Kline, Tashman, & Zauel, 2008; Bey, Zauel, Brock, & Tashman, 2006) Optical-based motion capture systems exhibit an inherent error in their measurements due to soft tissue artifacts which occur as the surrounding tissue moves with respect to the joint being studied.(Gale & Anderst, 2019) More exact methods of quantifying joint motion such as fixing the imaged markers to the bone using intracortical pins are extremely invasive, costly, and not feasible for frequent analysis.(Bey et al., 2008; Bey et al., 2006; Gale & Anderst, 2019) To analyze joint motion with six degrees of freedom (translation in three directions and rotations about three axes) biplane radiography provides a reliable and accurate method of capturing bone motion *in-vivo* during dynamic movements.(Anderst, Zauel, Bishop, Demps, & Tashman, 2009; Bey et al., 2006)

2.3.6 Methodological Considerations

Accurately analyzing joint function requires sophisticated data analysis methods that minimize the effects of soft tissue artifact that are evident in optical marker-based skin motion capture analysis. Biplane radiography and model-based tracking allows for the measurement of *in-vivo* joint motion by matching a subject-specific CT-generated volumetric model to radiographic images acquired during data collection of a dynamic movement such as running or walking. To quantify tibiofemoral joint kinematics, subject-specific three-dimensional models of the distal femur and proximal tibia are created using Mimics segmentation software (Materialise, Leuven, Belgium). The model-based tracking technique tracks position of bones based on their 3D shape and texture without requiring surgically implanted beads and has been validated in various joints such as the glenohumeral joint, the tibiofemoral joint, the ankle, and the patellofemoral joint, to name a few. (Anderst et al., 2009; Bey et al., 2008; Bey et al., 2006; Iaquinto et al., 2018) To register the collected biplane radiographs with the subject-specific 3D model, a validated model-based tracking algorithm is used to match the model position to the bone position within an individual frame of the radiograph.(Anderst et al., 2009) Once the model's position has been established in at least two frames, the automated tracking algorithm interpolates the model's position throughout all frames of the radiograph data. This automated optimization process can then be refined and run again to create an accurate representation of joint movement. This technology is discussed in detail with an explanation of data processing workflow in the Data Analysis section.

When conducting kinematic analyses using biplane radiography it is important to minimize the amount of radiation the research subject is exposed to over the course of the study. Considerations for analyzing knee joint kinematics should address the necessity of imaging both limbs or, if only a single limb is sufficient, if limb dominance plays an important role in choosing which side to image. Previous studies have identified small side-to-side differences in knee joint kinematics when studying both limbs during walking.(Gale & Anderst, 2019) In a study of 19 healthy subjects performing a walking task on a treadmill at a self-selected speed, Gale et al. identified that differences in knee kinematics between the two limbs averaged 1.3 mm or less in translation and 3.8˚ or less in rotation over the complete gait cycle, with maximum differences of occurring during late stance phase (2.2 mm translation, 7.1˚ rotation).(Gale &

Anderst, 2019) Although these absolute side-to-side differences were significantly smaller for anterior-posterior translation and abduction-adduction rotation during stance phase than during swing phase, no effect of limb dominance on knee kinematic asymmetry was identified.(Gale & Anderst, 2019) In addition to providing a useful clinical tool for evaluating return to function in a pathological limb following clinical intervention, this research provides rationale for imaging a single limb when utilizing biplane radiography to assess knee motion in healthy subjects. In the absence of significant asymmetry, analyzing a single limb will limit both data collection time and the subject's exposure to radiation when utilizing a biplane X-ray system.

3.0 Methods

3.1 Experimental Design

This study utilizes a within-subject laboratory-based study design. The purpose of this study is to investigate the effects of load carriage and gait velocity on tibiofemoral joint motion, contact path, subchondral joint space, and knee joint kinematics in recruit-aged females.

3.2 Participants

3.2.1 Subject Recruitment

Study procedures were approved by the University of Pittsburgh's Institutional Review Board and study subjects were recruited from the University of Pittsburgh and the surrounding community. Recruitment flyers were posted in athletic facilities and academic buildings around University of Pittsburgh's' Oakland campus. Potential subjects contacted the Neuromuscular Research Laboratory (NMRL) for an initial pre-screening. If all eligibility criteria were met and consent was obtained, the subject was enrolled in the study and the first study visit was scheduled. Study sites included the University of Pittsburgh's Neuromuscular Research Laboratory (NMRL) and Orthopedic Biodynamics Laboratory (BDL).

3.2.2 Inclusion Criteria

Subjects were military recruit-aged females (aged 18-34). The study was limited to female subjects to provide a more heterogeneous population sample and in order to focus on the potential effects of load carriage on a female recruit population. Additionally, subjects were moderately physically active (30-60 minute of activity at least 5 times per week), comfortable wearing the tight clothing essential for accurate motion capture, and comfortable carrying loads up to 45% of their bodyweight.

3.2.3 Exclusion Criteria

Potential subjects were excluded from study participation if they met any of the following criteria:

- Current injury affecting their back or lower extremities
- Pre-existing conditions which would be worsened by participating in activities involving load carriage (i.e. history of back or lower extremity pain or injury.)
- Pregnant (if unknowingly pregnant, subject was ineligible to participate in study procedures due to the exposure to radiation and was removed from consideration following the pregnancy test at the Biodynamics Laboratory)

3.3 Sample Size Calculation

A sample of 12 women participated in the study. A power analysis conducted using G*Power version 3.1.9.2 (Heinrich Heine, Universität Dusseldorf, Germany) determined that 16 subjects were needed to have a power of 0.8 and effect size of 0.2 with a two-sided α of 0.05. To account for possible attrition and data loss, 18 subjects were targeted for study recruitment.

3.4 Instrumentation

3.4.1 Dual Energy X-ray Absorptiometry

A dual energy X-ray absorptiometry (DEXA) system (Lunar iDXA, GE Healthcare, Chicago, IL) was used to obtain body composition data from all study subjects. Height measurements for each subject were performed by study researchers using a stadiometer and entered into the DEXA system to ensure an accurate assessment of height. Weight was assessed using a standard scale. Percentage body fat, fat free mass, and fat mass were calculated as part of demographic data collection. Previous research has shown that DEXA is a valid methodology for analyzing body composition compared to methods such as hydrostatic weighing and CT-based measurements, and has been shown to reliably measure compartmental body composition in weight-stable adults (less than 6% variation over three months).(Dordevic et al., 2018; Haarbo, Gotfredsen, Hassager, & Christiansen, 1991; Hind, Oldroyd, & Truscott, 2011; Kaul et al., 2012; Rothney et al., 2012)

3.4.2 Vicon Motion Capture System

Kinematic data were captured using a Vicon Vantage (Vicon Motion Systems, Oxford, United Kingdom) system incorporating twelve cameras with a capture rate of 100 Hz. Reflective markers placed on the study subject's lower extremities and shoulders allowed for data capture of lower extremity movement and limited torso movement. When compared to joint motion calculated from reflective markers placed on intracortical bone pins fastened to the femur, motion capture using skin reflective markers have been shown to describe rotational and translational motion at the knee joint with errors of up to 4.48˚ and 13.0 mm respectively.(Benoit et al., 2006) All data were exported into Visual 3D (C-Motion Inc., Germantown, MD) for analysis. Marker placement for the study procedures is described in detail in later sections.

3.4.3 Biplane Radiography System

The University of Pittsburgh's Orthopedic Biodynamics Laboratory (BDL) houses a custom biplane radiography system with the ability to collect high-speed images during dynamic movements. Mounted on a gantry, the dual X-ray tubes can be reconfigured and collect images during a wide variety of movements at various joints. The system utilizes two 150 kVp constant-potential high-frequency cardiac cine radiographic generators (EMD CPX-3100 CV, EMD Technologies, Saint-Eustache, Quebec, Canada) which use short-duration pulses at high repetition rates to achieve high quality images of dynamic movements with minimal motion blur with an accuracy on the order of 1 mm in translation and 1˚in rotation.(Anderst et al., 2009; Bey et al., 2008) The biplane radiography system is synchronized with the Vicon motion capture system and the Bertec instrumented treadmill to allow for simultaneous data collection during dynamic movements.

3.4.4 Bertec Fully-Instrumented Split Belt Treadmill

A Bertec Fully-Instrumented Split Belt Treadmill (Columbus, Ohio) was utilized for its precise control over running velocity and treadmill belt acceleration during data collection. The Bertec treadmill incorporates a dual belt system with a separate force plate under each belt to allow for the collection of ground reaction forces under each foot. The treadmill was synchronized with both the biplane radiography system and the Vicon motion capture system to allow for simultaneous collection of motion capture data, biplane x-rays, and ground reaction forces.

3.5 Testing Procedures

All research procedures took place at the Neuromuscular Research Laboratory (NMRL) and Orthopedic Biodynamics Laboratory (BDL) within the University of Pittsburgh. Potential subjects called the NMRL for eligibility screening. Once the subject was deemed eligible, informed consent was obtained and the subject was enrolled in the study. The study procedures incorporated four total study visits: one orientation visit to the NMRL, one CT scan, and two data collection visits to the BDL. The first study visit involved final subject recruitment and consent, anthropomorphic measurements, and a shortened load carriage protocol to familiarize the subject with study procedures and interacting with the treadmill and load carriage system.

The second study visit was the first data collection day at the BDL, where the subject performed trials under all load carriage conditions. Radiographic images of the right knee were collected during this visit. Data collection was repeated in the same manner for an additional data collection day. A detailed overview of study procedures by visit is presented in the following sections.

3.5.1 Orientation Visit

Potential subjects reported to the NMRL for final study eligibility screening. If all inclusion criteria were met and consent was obtained, the subject was enrolled in the study. A urine pregnancy test was performed prior to data collection due to radiation exposure from the DEXA system. The scanner was calibrated before each use using the standard calibration block and manufacturer's instructions. Anthropomorphic and body composition measurements were obtained using DEXA, a standard scale to measure weight, and a stadiometer to measure height. Next, the subject completed a shortened protocol that allowed them to become familiar with the study procedures as described in section 3.5.4. For each loading condition, subjects walked on a treadmill for approximately 30 seconds before accelerating to a run for approximately 15 seconds. Approximately 5 minutes of walk/run cycles were completed for each loading condition. This protocol allowed the subject to become familiar with the combat boots, weighted vest, and treadmill before performing the full-length procedure at the BDL.

3.5.2 CT Scan

High-resolution computed tomography (CT) scans of each subjects' right knee were obtained at UPMC Mercy, burned to a disk by the radiologist, and transported to the BDL by a member of the research staff. A phantom was placed under the subject's right leg during the scan to allow for bone density correction prior to data processing. The scan dimensions were 125 x 125 x 0.5 mm. The scan was centered on the subject's knee joint line, allowing for approximately 125 mm of femoral bony tissue and 125 mm of tibial bony tissue to be imaged in 0.5 mm slices. Additionally, hip and ankle slices were obtained to aid in defining the shank and thigh coordinate systems for kinematic analysis. CT scans with identifiable information were kept in a locked cabinet at the BDL, per BDL protocol. All CT data were de-identified and bone density-corrected using a custom MATLAB script prior to processing.

3.5.3 Lab Configuration

Each subject completed two data collection visits. Prior to the subject's arrival at the BDL, a subject file was set up in the Vicon system software and the system was calibrated. The X-ray sources and detectors were also positioned in the approximate configuration to analyze the right knee joint. The offset emitter and image intensifier captured the knee in the anterior/posterior direction while the inline emitter and image intensifier captured an oblique view of the same knee, with an 1800 mm source-to-detector distance and no incline or decline. This configuration allowed for the least occlusion from the contralateral leg. The angle between the inline and offset image intensifiers was 55˚. No magnification or filtering was used. A hand rail with emergency stop button was positioned on the subject's right side, and subjects were instructed to press the button if they felt unsteady at any point during data collection. Final adjustments to the height of the biplane radiography system position were made once the subject entered the lab and positioned themselves on the treadmill.

The 12-camera Vicon system was configured with 8 cameras mounted around the perimeter of the lab space, one camera directly above the subject, two cameras capturing a lateral and oblique view to the subject's left and one camera capturing a lateral view to the subject's right. After positioning the Vicon cameras, masking and calibration with a wand were performed. The lab coordinate system was established by placing a wand with reflective markers on the center of the treadmill. While on the treadmill, the positive x-direction pointed laterally to the subject's right, the positive-y direction pointed anteriorly, and the positive-z direction pointed superiorly. Force plate data were collected at 1000 Hz. A calibration cube was utilized to determine the laboratory coordinate system and allow for recreation of the laboratory space during the model-based tracking process, as well as to precisely determine the locations of the x-ray sources and image intensifiers during the data collection process.

3.5.4 Data Collection Visits

For each data collection visit, subjects reported to the BDL. Subjects were required to wear compression shorts and tank tops, as well as a swim cap to hold the hair away from the neck to prevent occlusion of the neck and torso markers. Combat boots were provided by the research staff, but subjects were encouraged to wear high socks to increase the comfort of the boots. Reflective markers were then placed on select anatomical landmarks to establish the measured and anatomical coordinate systems of the femur and tibia. Markers were placed bilaterally at the following locations:

- Superior surface of acromion
- 1st thoracic vertebrae
- Anterior superior iliac spine (ASIS)
- Posterior superior iliac spine (PSIS)
- Greater trochanter of the femur
- Medial and lateral femoral condyles
- Fibular head
- Thigh cluster
- Shank cluster
- Medial and lateral malleolus
- 1st metatarsal
- Medial and lateral MP joint of foot

Once the markers were placed, the subject entered the testing room and positioned themselves on the treadmill, and both static and dynamic calibrations were performed.

One researcher was present in the lab space with the subject during the testing while another researcher operated the Vicon and biplane radiography system from a control room separated from the x-ray space by a leaded glass window to minimize radiation exposure. The researcher interacting with the subject wore proper x-ray protective equipment (lead vest and dosimeter badge) and served to instruct and reposition the subject during data collection trials and trigger the acquisition of simultaneous radiographic and motion capture images. Load and velocity conditions were randomized. All load carriage conditions were accomplished using a weighted vest with the load divided evenly anteriorly and posteriorly. For each load condition (body weight (BW), +25% BW, and +45% BW) the subject completed a gait transition speed protocol and three trials of data collection with x-ray exposure for a total of nine trials.

Three trials were completed in a randomized fashion for each load condition. For each load condition the subject's transition velocity was determined by asking the subject to begin walking on the treadmill with a constant acceleration of 0.05 m/s^2 and naturally transition to a run. The researchers recorded the speed at which the transition to running occurred and used an average of three trials to determine the subject's average transition velocity for that loading condition. No data was collected during these trials. Using this transition velocity (TV) three trials of data collection were performed per loading condition. All trials followed a stepped protocol of one minute of locomotion at a lower speed, an acceleration of 0.05 m/s^2 to reach the higher speed, and approximately one minute of locomotion held at the higher speed. Subjects were instructed to run if necessary for two out of the three trials, but for the third trial they were instructed to maintain a walking gait even if they felt as though transitioning to running would be more optimal. The three trials of data collection are outlined below:

1. 1 minute at 10% below TV, 1 minute at 10% above TV, subject transitioned to run if necessary
2. 1 min at 30% below TV, 1 minute at 10% below TV, subject transitioned to run if necessary
3. 1 minute at 10% below TV, 1 minute at 10% above TV, subject instructed not to transition to a run ("forced march" condition)

The researcher repositioned the subject as necessary to ensure the right knee remained in the system field of view. Synchronized biplane radiographs (150 images/sec for 1.0s, maximum 90 kV, 160 mA, 1 ms pulse width) of the knee were collected at right heel strike for all dynamic trials. Static radiographs were captured (0.5 s duration) for all loading conditions. Subjects were given a rest period between trials and load conditions to minimize fatigue. Upon completion of the study visit all data was exported for analysis. The study procedures were completed for an additional study visit. Subjects were compensated based on their completion of study procedures with all payments loaded onto a personal card via the University of Pittsburgh's online payment system. The direction of the lab-based coordinate system and location of the x-ray sources and detectors was established by taking a radiograph of a calibration cube.(Anderst et al., 2009) These parameters were used to recreate an identical test configuration in virtual space for the model-based tracking of the femur and tibia.(Anderst et al., 2009)

3.6 Data Reduction

No magnification or filtering were used during the collection of radiographic images. The overall workflow utilized to analyze joint kinematics is illustrated in Figure 1 and described in detail in the following sections.

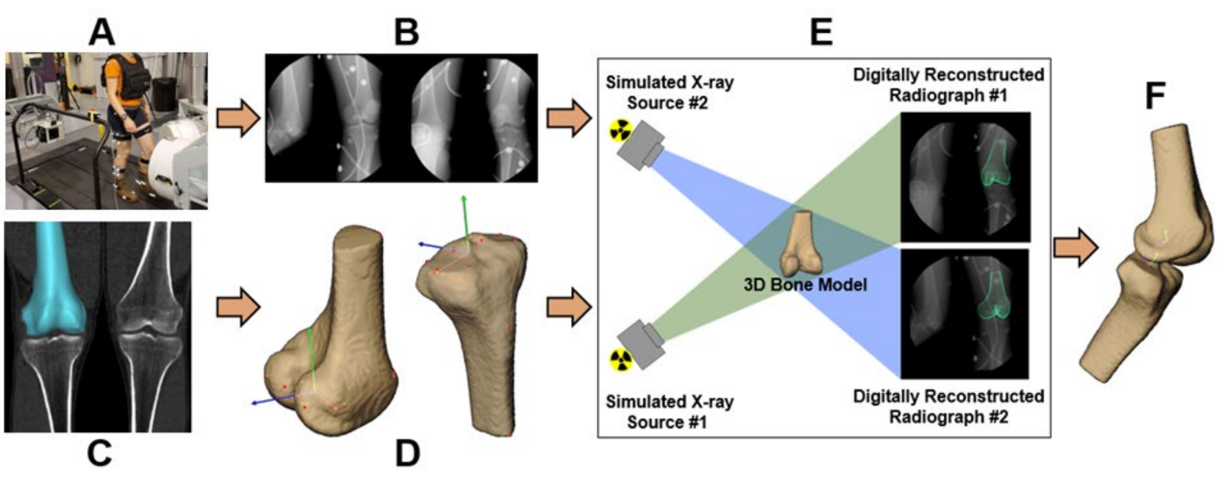

Figure 1: Data Processing Workflow

The workflow utilized to acquire and analyze biplane radiographic images of dynamic tibiofemoral joint motion. (A) Subject performs data collection trials on treadmill resulting in approximately 150 frames of x-ray data from two viewing angles, as shown in (B). (C) Subject CT scans are segmented to create subject-specific 3D bone models with coordinate systems established in (D). (E) 3D bone models are matched to the radiographic images in the virtual lab space using a model-based tracking technique. (F) Six DOF kinematics of the tibiofemoral joint are generated for further analysis.

3.6.1 Creating Bone Models

Three-dimensional subject-specific models of the femur and tibia were created for digital reconstruction of the x-ray system during the model-based tracking process.

3.6.1.1 CT Segmentation

Subject CT scans were anonymized using a custom MATLAB script for data extraction and storage on the BDL servers. All CT data was received in DICOM format. Disks with identifiable information were stored in a locked cabinet, as per BDL procedure. Bone density-corrected sagittal reconstructed images of the knee joint were imported into Mimics Version 20 (Materialise, Leuven, Belgium). This software uses the DICOM images obtained from the CT scanner and a set of segmentation tools to aid the user in isolating the bone in question from the full CT scan and export the data into a variety of different formats. Besides its utility in creating volumetric bone models necessary for the model-based tracking process, the Mimics analysis suite has research applications in surgical planning, patient-specific device design, and taking anatomical measurements in a non-invasive manner. Before beginning segmentation, the CT images were resliced to the correct image dimension and pixel size. The Mimics masking tool was used to isolate femoral and tibial tissue from the full CT images in the sagittal view based on a thresholding tool which allowed for the identification of the bony perimeter in each sagittal slice. Manual manipulation of the mask was performed in order to ensure accurate detection of bony tissue. The resulting CT volumes included the outer cortical bone surface and the interior bony tissue. Smoothed masks of the femur and tibia were exported to BMP/JPEG format and used as a template to isolate bony tissue from the full CT images. 3D bone models were created using a custom software script and were used in subsequent model-based tracking steps.

3.6.1.2 Surface Creation

Using a custom software program provided by the Biodynamics Laboratory, 3D models of the femur and tibia were generated using exported masks of each subject's CT scan. This software allows the user to place a marker on anatomical landmarks to define a coordinate system for each bone that was imaged using biplane radiography. Landmarks were placed on the anterior, posterior, medial, and lateral borders of the tibial plateau to define the boundaries of the medial and lateral tibial plateau. Regions of interest (ROIs) were fit to these points to allow for the analysis of contact path, contact area, and joint space within different subregions of the plateau. Anatomical coordinate systems were established for the femur and tibia using a previously established method illustrated in Figures 2.(Anderst et al., 2009; Gale & Anderst, 2019) For the femur, the anatomical x-axis was defined as the vector connecting the condylar centers pointing laterally. The anatomical z-axis was then defined as the cross product of the anatomical x-axis and the vector connecting the average of the condylar centers and the hip joint center, and the anatomical y-axis was the cross product of the anatomical z- and x-axes. For the tibia, the anatomical y-axis was defined as the vector from the ankle joint center to the point between the tibial condyles. The anatomical z-axis was then the cross product of the anatomical y-axis and the vector connecting the lateral compartment to the medial compartment, and the anatomical x-axis was the cross product of the anatomical y- and z-axes.

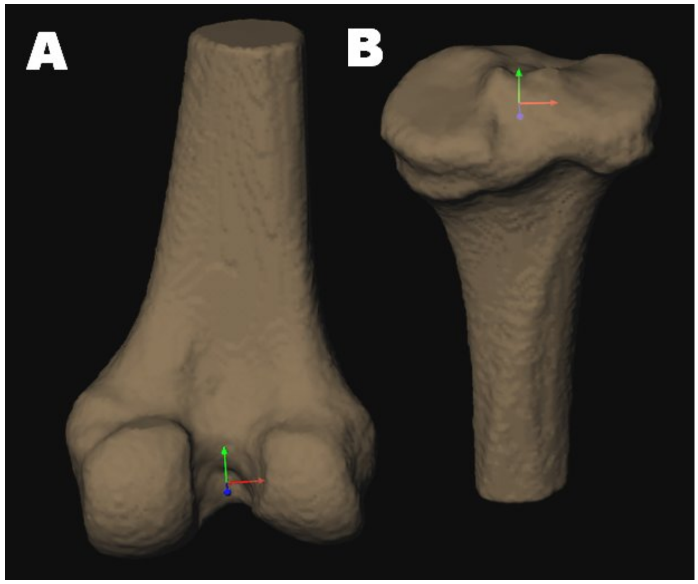

Figure 2: Anatomical coordinate systems for the femur (A) and tibia (B) as described by Gale et al.(Gale & Anderst, 2019)

3.6.2 Model-Based Tracking

Following data collection, radiographic images were exported to a disk and transferred to the BDL servers for analysis. All images were distortion corrected and bead tracked prior to beginning the model-based tracking process. A previously validated volumetric model-based tracking technique with an *in-vivo* accuracy of 0.7 mm or better in translation and 0.9˚ or better in rotation was utilized to match the model position to the bone position within an individual frame of the radiograph.(Anderst et al., 2009; Bey et al., 2008) This algorithm maximizes the correlation between the biplane radiographs and the digitally reconstructed radiographs. Once the model's position has been established in at least two frames, the automated tracking algorithm interpolates and optimizes the model's position throughout all frames of the radiograph data. This automated optimization process can then be refined and run again to create an accurate representation of joint movement. Tracking was performed for the femur and tibia for each movement trial, provided clear trackable images were obtained. A non-trackable image included image sequences where either the tibia or femur was not within the imaging frame (either by error in system positioning or by the subject moving outside the imaging area) resulting in an incomplete collection of stance phase data. If model-based tracking software was unable to create an accurate representation of the joint movement, the data for that trial was excluded from analysis. Nine trials were tracked for each subject to account for each load and velocity combination, as well as a static trial for each loading condition. It also should be noted that smoothing and interpolation of bone movements is an integral part of the model-based tracking process, so this should be accounted for when considering the outcomes of the study. Tracking was considered complete when the 3D bone model matched the radiographs in each frame with no obvious algorithmic artifacts influencing the bone placement. Each bone's position and orientation in each frame (X, Y, Z and roll, pitch, yaw) was used for kinematic analysis.

3.6.3 Kinematic Analysis

Tibiofemoral rotational and translational kinematics were calculated using the model-based tracking solutions for each trial (flexion/extension, abduction/adduction, and internal/external rotations and anterior/posterior, medial/lateral, and proximal/distal translations). To analyze the arthrokinematics of the knee, axes of rotation were established as previously with flexion/extension about the femoral medial/lateral axis, internal/external rotation about the proximal/distal tibial axis, and abduction/adduction about a floating axis, illustrated in Figure 3.(Grood & Suntay, 1983) Translations were calculated from the femoral anatomic origin to the tibial anatomic origin and expressed in the tibial anatomic coordinate system.(Gale & Anderst, 2019) Right foot strike was identified using the ground reaction forces recorded by the instrumented treadmill to signify the initiation of the stance phase of gait. The 6 DOF kinematics were interpolated to percent gait cycle. Knee kinematics were analyzed at initiation of support phase (right foot strike) and at 10% increments for the amount of support phase captured by the biplane radiography system. Contact path was determined by the kinematics software by identifying the location center of closest contact between the subchondral femur and tibia within the medial and lateral compartments for each frame. Center of closest contact locations were compared between load and velocity conditions. Dynamic joint space was also determined by the kinematics software using the overall amount of space present between the subchondral bone surfaces. Average joint space and minimum joint space were analyzed at the same time points used to analyze tibiofemoral kinematics.

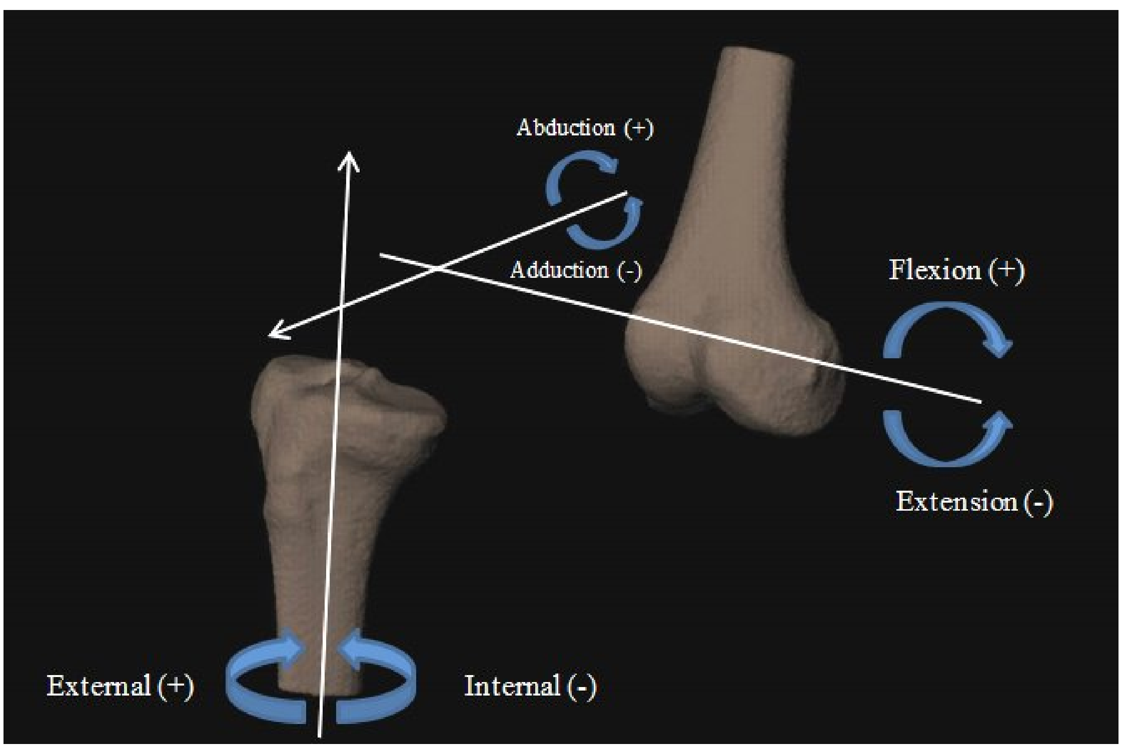

Figure 3: Joint axes of rotation as described by Grood et al.(Grood & Suntay, 1983)

3.7 Statistical Analysis

All data was analyzed using SPSS Version 26 (IBM Inc.; Armonk, NY). Descriptive statistics (means, standard deviations, medians, and interquartile ranges as appropriate) were calculated for all variables. Shapiro-Wilk tests were used to assess normality for all dependent variables. Separate two-way (Load X Locomotion) repeated measures analysis of variance were used to examine the effect of load (3 repeats; BW, +25%BW, +45%BW) and locomotion (3 repeats; walk, run, and force march) tibiofemoral kinematic characteristics. For data that was not normally distributed, corresponding non-parametric tests were used. Alpha was set *a priori* at 0.05, two-sided.

4.0 Results

The purpose of this study was to investigate the effects of load carriage magnitude and gait velocity on tibiofemoral arthrokinematics in a military recruit-age female population. The effects of load carriage magnitude relative to bodyweight and the effects of gait velocity on these biomechanical variables were analyzed separately, as well as their interactive effects. All variables were found to be normally distributed following the Shapiro-Wilk analysis for normality ($p>0.05$), allowing for the use of parametric tests during data analysis.

4.1 Descriptive Data

4.1.1 Demographic Data

Age, height, weight, and relative load amounts for the analyzed load carriage conditions are reported in Table 1.

Table 1: Descriptive data and load carriage magnitudes.

	N	Mean ± SD	Median	IQR (Q1, Q3)
Age (years)	12	24.5 ± 2.4	23.3	23.3, 27.0
Height (cm)	12	161.7 ± 5.7	163.4	157.2, 165.4
Body Mass (kg)	12	58.1 ± 7.8	57.7	51.9, 62.1
+25%BW Load Mass (kg)	12	14.3 ± 2.0	14.4	12.5, 15.1
+45%BW Load Mass (kg)	12	25.6 ±3.5	25.2	23.1, 27.2
BMI (kg/m^2)	12	22.2 ± 2.3	21.6	20.4, 23.2
Fat Free Mass (kg)	12	42.8 ± 6.1	41.6	39.8, 43.3
Percent Fat Free Mass (%)	12	71.0 ± 8.4	70.4	64.4, 76.0
Fat Mass (kg)	12	15.6 ± 6.7	16.3	11.1, 19.7
Percent Fat Mass (%)	12	25.9 ± 9.2	26.7	20.5, 33.1

4.1.2 The Interactive Effects of Load Carriage Magnitude and Gait Velocity

Tibiofemoral kinematics were analyzed from right heel strike to 30% of right leg support at intervals of 10%. Analysis was limited to 30% of right leg support due to the common range of available data across all subjects. Minimum gap distance of the medial and lateral compartments of the knee were analyzed at the same time points. Contact path length over the common range (right heel strike to 30% right leg support) was analyzed for the medial and lateral compartments of the knee.

A 3x3 two-way repeated measures analysis of variance was performed on each of the six DOF kinematics (flexion, abduction, internal rotation, medial translation, proximal translation, and anterior translation), medial and lateral compartment minimum gap, and medial and lateral contact path length as a function of load and locomotion. The within-subjects independent variable load had 3 levels (BW, +25BW%, +45BW%) and the within-subjects independent variable locomotion had 3 levels (walk, forced march, run). In order to find the pattern of differences in kinematics averaged across load or locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. The results for flexion, medial and lateral compartment minimum gap, and medial and lateral contact path length are presented in this section, while results for abduction, internal rotation, medial translation, proximal translation, and anterior translation can be found in the Appendix.

4.1.2.1 Tibiofemoral Kinematics at Right Heel Strike

Average tibiofemoral kinematics at right heel strike are reported in Table 2.

Table 2: Average tibiofemoral kinematics at right heel strike (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Flexion (degrees)	Walk	BW	4.9 ±9.1	2.8	0.4, 5.1
		+25%BW	10.5 ± 10.1	10.1	4.9, 15.9
		+45%BW	11.4 ± 7.9	12.5	5.8, 17.5
	Force March	BW	14.4 ± 8.9	15.2	10.7 18.8
		+25%BW	18.7 ± 12.1	16.8	12.7, 24.5
		+45%BW	16.4 ± 9.9	16.9	11.3, 25.1
	Run	BW	15.4 ± 7.5	13.4	11.1, 18.0
		+25%BW	15.9 ± 7.1	15.5	10.4, 19.5
		+45%BW	17.4 ± 4.6	18.7	14.2, 20.5
Abduction (degrees)	Walk	BW	-3.2 ± 4.0	-3.7	-5.9, -0.1
		+25%BW	-3.9 ± 4.0	-4.1	-5.4, -2.7
		+45%BW	-4.4± 2.9	-4.5	-5.7, -2.8
	Force March	BW	-4.0 ± 4.2	-4.3	-6.8, -1.1
		+25%BW	-4.5 ± 3.6	-4.9	-7.4, -2.4
		+45%BW	-4.3 ± 3.5	-4	-6.3, -1.2
	Run	BW	-3.9 ± 3.4	-4.0	-5.7, -1.3
		+25%BW	-4.4 ± 3.7	-4.5	-6.1, 1.9
		+45%BW	-4.4 ±3.4	-4.6	-6.3, -1.8
Internal Rotation (degrees)	Walk	BW	-0.5 ± 7.3	0.3	-3.7, 3.1
		+25%BW	-1.5 ± 7.2	-2.7	-4.8, 1.0
		+45%BW	-1.7± 7.8	-3.4	-6.1, 1.2
	Force March	BW	0.3 ± 8.4	-0.4	-2.3, 2.1
		+25%BW	-1.0 ± 7.8	-0.9	-5.8, 2.6
		+45%BW	-0.1, 8.3	-0.9	-4.8, 0.8
	Run	BW	-0.7 ± 7.6	-0.4	-4.1, 2.8
		+25%BW	0.5 ± 9.2	0.5	-5.2, 2.4
		+45%BW	0.7 ± 8.1	0.5	-3.2, 3.3

Table 2 (continued)

Medial Translation (mm)	**Walk**	**BW**	3.1 ± 1.6	3	2.0, 4.4
		+25%BW	2.4 ± 1.7	2.5	0.7, 3.4
		+45%BW	2.6 ± 1.7	2.3	1.5, 3.5
	Force March	**BW**	2.5 ± 1.7	2.3	1.5, 3.9
		+25%BW	2.5 ± 1.5	2.2	1.5, 3.2
		+45%BW	2.5 ± 1.9	1.8	1.1, 3.9
	Run	**BW**	2.7 ± 1.5	3.1	1.3, 3.6
		+25%BW	2.9 ± 1.8	2.8	1.8, 4.1
		+45%BW	2.8 ± 1.2	3	1.9, 3.7
Proximal Translation (mm)	**Walk**	**BW**	-24.5 ± 2.6	-25.1	-26.2, -22.5
		+25%BW	-23.4 ± 3.0	-24.5	-25.4, -21.7
		+45%BW	-23.0, 2.0	-23.5	-24.2, -21.6
	Force March	**BW**	-22.0 ± 2.9	-21.3	-24.1, -20.9
		+25%BW	-21.8 ± 2.8	-22.5	-23.8, -19.5
		+45%BW	-22.5 ± 2.3	-23.0	-24.7, -20.9
	Run	**BW**	-23.3, 2.6	-24.3	-25.4, 21.5
		+25%BW	-23.1 ± 1.8	-22.9	-24.8, -21.5
		+45%BW	-22.8 ± 1.6	-23.3	-23.9, -21.4
Anterior Translation (mm)	**Walk**	**BW**	6.3 ± 3.2	6.1	5.0, 7.5
		+25%BW	5.5 ± 3.0	6.1	4.2, 7.5
		+45%BW	5.4 ±3.4	6	4.2, 7.5
	Force March	**BW**	5.2 ± 2.6	4.7	4.1, 6.2
		+25%BW	4.5 ± 2.6	4.3	2.7, 5.8
		+45%BW	4.5 ± 2.6	3.8	3.1, 6.5
	Run	**BW**	5.1 ± 3.2	4.7	3.2, 7.7
		+25%BW	4.6 ± 2.4	4.8	3.6, 6.2
		+45%BW	4.7 ± 2.7	4.2	3.7, 7.1

There was no significant interaction between load and locomotion in their effect on flexion ($F_{4,44} = 0.641$, $p = 0.636$, $\eta_p^2 = 0.55$). There was a significant main effect of locomotion on flexion, averaged across levels of load ($F_{2,22} = 10.007$, $p = 0.001$, $\eta_p^2 = 0.476$) (Figure 4). Flexion was significantly lower at the walking category of locomotion (mean = 8.9˚, SE = 9.5˚) than at the forced marching category of locomotion (mean = 16.5˚, SE = 10.5˚), averaged across levels of load (p = 0.010). Flexion was significantly lower at the walking category of locomotion (mean = 8.9˚, SE = 9.5˚) than at the running category of locomotion (mean = 16.2˚, SE = 6.6˚), averaged across levels of load (p = 0.010). There was no significant main effect of load on flexion, averaged across locomotion.

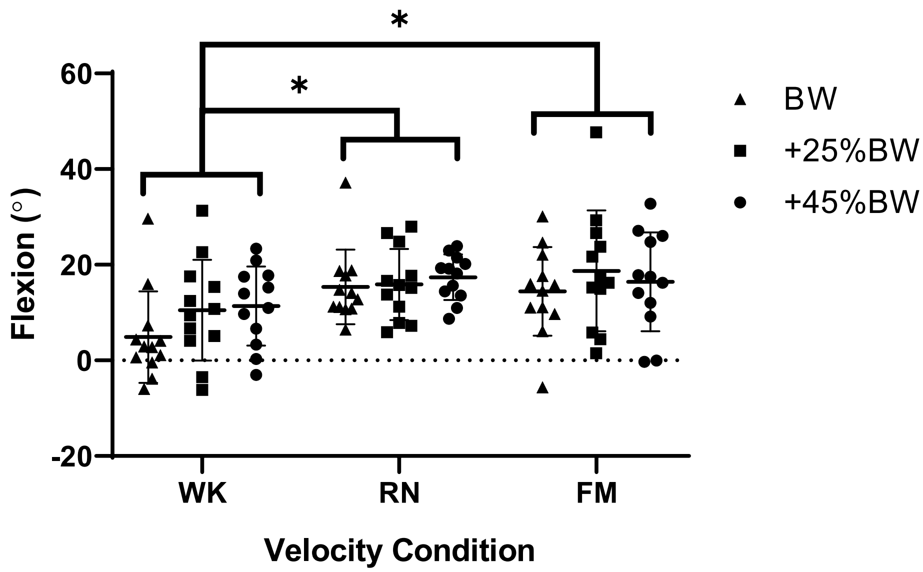

Figure 4: Average flexion at right heel strike.

Flexion angle (˚) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

4.1.2.2 Tibiofemoral Kinematics at 10% Right Leg Support

Average tibiofemoral kinematics at 10% right leg support are presented in Table 3.

Table 3: Average tibiofemoral kinematics at 10% right leg support (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Flexion (degrees)	Walk	BW	3.4 ± 3.4	3.1	1.9, 4.3
		+25%BW	6.1 ± 5.4	8.1	0.4, 9.8
		+45%BW	6.3 ± 6.7	5.5	1.1, 8.3
	Force March	BW	6.8 ± 5.2	7.8	2.6, 10.7
		+25%BW	8.9 ±7.2	9.2	5.2, 12.8
		+45%BW	9.6 ± 7.3	10.1	4.4, 13.1
	Run	BW	15.7 ± 4.4	14.9	13.7, 18.2
		+25%BW	14.3 ± 8.3	15.7	12.6, 18.2
		+45%BW	15.6 ± 6.5	16.5	13.6, 20.1
Abduction (degrees)	Walk	BW	-3.1 ± 3.4	-3.7	-4.9, -0.3
		+25%BW	-3.7 ± 3.1	-4.1	-5.3, -2.7
		+45%BW	-3.8 ± 3.4	-4.6	-5.6, -2.6
	Force March	BW	-3.4 ± 3.6	-4.1	-5.6, -0.3
		+25%BW	-3.7 ± 3.9	-4.1	-6.2, -1.2
		+45%BW	-3.5 ± 3.8	-4.3	-5.6, -0.3
	Run	BW	-4.2 ± 3.6	-4.5	-6.1, -1.3
		+25%BW	-4.4 ±3.6	-4.6	-6.3, -1.3
		+45%BW	-4.4 ± 3.5	-4.6	-6.5, -1.6
Internal Rotation (degrees)	Walk	BW	0.3 ± 7.2	-0.1	-3.8, 2.6
		+25%BW	-0.8 ± 7.3	-0.7	-4.3, 3.4
		+45%BW	-0.9 ± 7.5	-1.2	-3.8, 2.6
	Force March	BW	0.0 ± 7.7	-1.2	-2.2, 8,8
		+25%BW	-0.1 ± 8.0	-1.5	-3.7, 4.1
		+45%BW	0.5 ± 8.0	-0.2	-2.4, 1.3
	Run	BW	0.9 ± 7.2	0.2	-2.0, 4.7
		+25%BW	1.2 ±7.2	0.0	-3.4, 4.6
		+45%BW	2.3 ± 7.7	0.6	-1.9, 6.6

Table 3 (continued)

Medial Translation (mm)	**Walk**	**BW**	3.9 ± 1.4	3.7	2.9, 5.1
		+25%BW	3.6 ± 1.6	3.4	2.4, 5.3
		+45%BW	3.4 ± 1.5	3.6	2.2, 4.6
	Force March	**BW**	3.5 ± 1.2	3.3	2.5, 4.6
		+25%BW	3.5 ± 1.7	3.7	1.9, 4.8
		+45%BW	3.4 ± 1.6	3.2	2.6, 4.7
	Run	**BW**	3.4 ± 1.4	3.8	2.1, 4.4
		+25%BW	3.5 ± 1.3	3.5	2.2, 4.6
		+45%BW	3.7 ± 1.4	3.8	2.1, 4.4
Proximal Translation (mm)	**Walk**	**BW**	-24.9 ± 2.2	-25.9	-26.6, -23.2
		+25%BW	-24.6 ± 2.3	-25.7	-26.3, -23.1
		+45%BW	-24.4 ± 1.8	-24.9	-26.3, -23.1
	Force March	**BW**	-24.5 ± 2.2	-25.3	-26.0, -22.8
		+25%BW	-24.2 ± 2.4	-25.3	-25.9, -22.4
		+45%BW	-23.9 ± 2.1	-25	-25.3, -21.9
	Run	**BW**	-22.8 ± 2.4	-24.1	24.7, -21.1
		+25%BW	-23.2 ± 1.9	-23.8	-24.8, -21.4
		+45%BW	-23.1 ± 2.1	-23.1	-24.7, -21.5
Anterior Translation (mm)	**Walk**	**BW**	7.4 ± 3.5	7.6	5.5, 8.9
		+25%BW	6.9 ± 3.4	6.9	5.0, 8.3
		+45%BW	7.1 ± 3.5	7.7	5.2, 9.4
	Force March	**BW**	6.0 ± 3.4	5.8	5.1, 7.0
		+25%BW	5.9 ± 2.6	6.2	4.8, 6.4
		+45%BW	6.2 ± 3.3	6.4	4.7, 8.0
	Run	**BW**	6.4 ± 3.8	6.2	3.0, 9.5
		+25%BW	6.1 ± 2.7	6.2	5.1, 7.8
		+45%BW	6.7 ± 3.0	7	5.6, 9.5

There was no significant interaction between load and locomotion in their effect on flexion ($F_{4,44} = 0.865$, $p = 0.493$, $\eta_p^2 = 0.073$). There was a significant main effect of locomotion on flexion averaged across levels of load ($F_{2,22} = 25.466$, $p < 0.001$, $\eta_p^2 = 0.706$) (Figure 5). Flexion was significantly lower at the walking level of locomotion (mean = 5.3˚, SE = 5.5˚) than the running level of locomotion (mean = 15.2˚, SE = 6.6˚), averaged across levels of load (p <0.001). Flexion was significantly lower at the forced marching level of locomotion (mean = 8.4˚, SE = 6.8˚) than the running level of locomotion (mean = 15.2˚, SE = 6.6˚), averaged across levels of load (p <0.001). There was no significant main effect of load on flexion, averaged across locomotion.

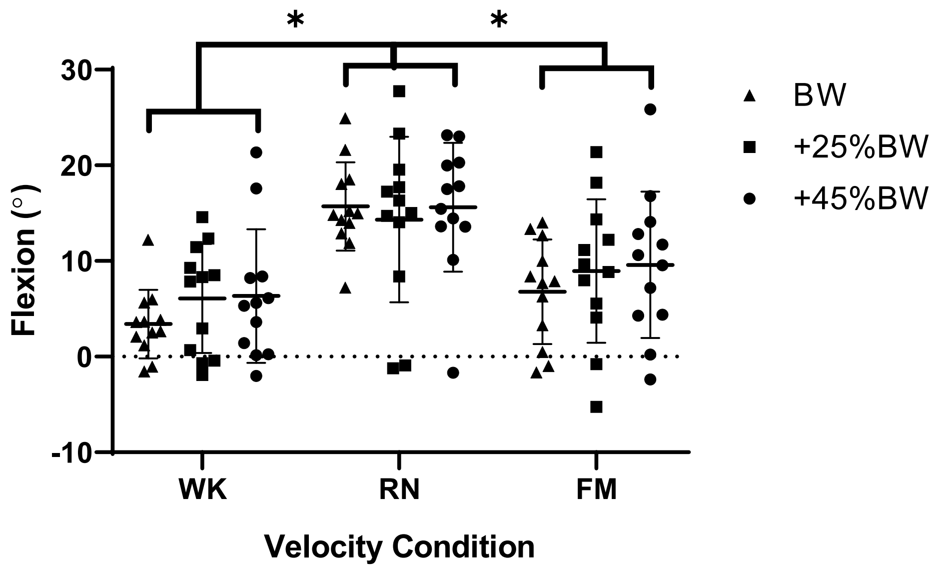

Figure 5: Average flexion at 10% right leg support.

Flexion angle (˚) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

4.1.2.3 Tibiofemoral Kinematics at 20% Right Leg Support

Average tibiofemoral kinematics at 20% right leg support are presented in Table 4.

Table 4: Average tibiofemoral kinematics at 20% right leg support (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Flexion (degrees)	**Walk**	**BW**	10.8 ± 4.5	9.7	8.3, 12.6
		+25%BW	12.6 ± 6.5	15.5	6.6, 18.3
		+45%BW	12.6 ± 5.3	11.8	10.2, 16.5
	Force March	**BW**	10.4 ± 3.7	11.1	7.3, 13.2
		+25%BW	12.6 ± 7.4	13.6	9.6, 14.5
		+45%BW	15.9 ± 7.4	15.0	11.5, 19.9
	Run	**BW**	20.4 ± 4.8	20.7	17.1, 23.3
		+25%BW	19.7 ± 8.0	21.1	17.5, 23.2
		+45%BW	21.4 ± 6.9	22.6	17.8, 26.4
Abduction (degrees)	**Walk**	**BW**	4.0 ± 3.3	-4.2	-6.5, -0.8
		+25%BW	-4.4 ± 3.1	-5.0	-6.7, -2.6
		+45%BW	-4.6 ± 3.6	-5.2	-6.6, -2.8
	Force March	**BW**	-4.0 ± 3.5	-4.5	-5.7, -0.8
		+25%BW	-4.2 ± 3.7	-4.3	-6.3, -1.3
		+45%BW	-4.1 ± 3.8	-4.2	-6.6, -1.2
	Run	**BW**	-4.8 ± 3.6	-4.6	-7.1, -1.7
		+25%BW	-4.8 ± 3.6	-4.2	-7.1, -1.6
		+45%BW	-4.7 ± 3.6	-4.0	-7.0, -1.7
Internal Rotation (degrees)	**Walk**	**BW**	1.8 ± 6.6	2.7	0.2, 5.0
		+25%BW	2.2 ± 7.2	2.7	-1.3, 6.7
		+45%BW	1.7 ± 6.6	1.5	-0.8, 5.0
	Force March	**BW**	2.2 ± 7.0	2.0	-1.3, 4.6
		+25%BW	2.3 ± 6.9	2.5	-0.4, 5.8
		+45%BW	4.4 ± 7.2	2.7	2.3, 6.9
	Run	**BW**	3.5 ± 6.6	2.8	0.9, 6.5
		+25%BW	3.9 ± 6.3	2.1	-0.7, 8.1
		+45%BW	5.8 ±6.9	4.5	1.4, 10.6

Table 4 (continued)

Medial Translation (mm)	**Walk**	**BW**	4.2 ± 1.4	4.9	3.1, 5.4
		+25%BW	4.0 ± 1.3	3.8	3.2, 5.1
		+45%BW	3.8 ± 1.5	3.9	2.4, 5.0
	Force March	**BW**	4.6 ± 1.4	4.9	4.0, 5.7
		+25%BW	4.2 ± 1.4	4.3	3.5, 5.1
		+45%BW	4.4 ± 1.4	5.1	3.7, 5.4
	Run	**BW**	4.0 ± 1.4	4.2	2.8, 5.1
		+25%BW	4.0 ± 1.3	4.2	2.7, 5.0
		+45%BW	4.3 ± 1.6	4.6	2.8, 5.2
Proximal Translation (mm)	**Walk**	**BW**	-23.0 ± 2.6	-24.1	-25.1, -21.4
		+25%BW	-22.8 ± 2.2	-23.3	-24.4, -21.5
		+45%BW	-22.8 ± 2.0	-23.2	-24.0, -21.1
	Force March	**BW**	-23.4 ± 2.0	-23.8	-24.7, -22.5
		+25%BW	-22.9 ± 2.0	-23.7	-24.4, -21.2
		+45%BW	-22.1 ± 2.6	-23.2	-23.9, -19.3
	Run	**BW**	-21.6 ± 2.2	-22.0	-23.1, -19.8
		+25%BW	-21.6 ± 2.0	-21.8	-23.1, -19.7
		+45%BW	-21.3 ± 2.2	-21.3	-23.0, -19.7
Anterior Translation (mm)	**Walk**	**BW**	9.1 ± 3.6	9.1	7.3, 111.4
		+25%BW	9.4 ± 3.8	8.8	7.6, 11.9
		+45%BW	9.2 ± 3.7	9.0	7.9, 11.5
	Force March	**BW**	8.2 ± 3.8	7.8	6.4, 10.0
		+25%BW	8.6 ± 3.4	8.7	7.5, 9.5
		+45%BW	9.4 ± 3.9	9.7	7.6, 11.4
	Run	**BW**	8.2 ± 4.3	8.4	3.6, 11.9
		+25%BW	8.6 ± 3.3	8.3	7.8, 10.3
		+45%BW	9.6 ± 3.5	9.9	8.0, 11.3

There was no significant interaction between load and locomotion, in their effect on flexion ($F_{4,44} = 1.072$, $p = 0.382$, $\eta_p^2 = 0.089$). There was a significant main effect of locomotion on flexion averaged across levels of load ($F_{2,22} = 25.411$, $p = <.001$, $\eta_p^2 = 0.698$) (Figure 6). Flexion was significantly lower at the walking category of locomotion (mean = 12.0˚, SE = 5.6˚) than the running category of locomotion (mean = 20.5˚, SE = 6.7˚), averaged across averaged across levels of load (p <0.001). There was no significant main effect of load on flexion, averaged across locomotion.

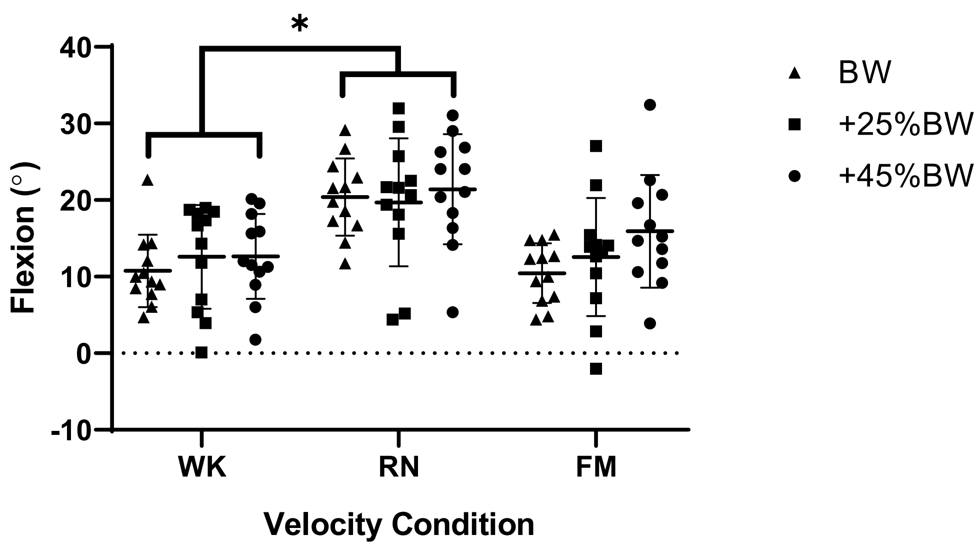

Figure 6: Average flexion at 20% right leg support.

Flexion angle (˚) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

4.1.2.4 Tibiofemoral Kinematics at 30% Right Leg Support

Average tibiofemoral kinematic data at 30% right leg support is presented in Table 5.

Table 5: Average tibiofemoral kinematics at 30% right leg support (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Flexion (degrees)	Walk	BW	17.2 ± 7.7	16.9	12.2, 17.9
		+25%BW	18.4 ± 6.8	20	13.2, 24.3
		+45%BW	17.4 ± 5.0	16.9	15.1, 21.9
	Force March	BW	17.6 ± 3.7	18.3	16.1, 20.2
		+25%BW	18.7 ± 6.2	19.2	16.0, 21.6
		+45%BW	22.4 ± 7.3	19.9	18.0, 24.8
	Run	BW	27.6 ± 6.5	28.4	25.1, 31.1
		+25%BW	27.0 ± 7.5	28.4	23.3, 30.0
		+45%BW	28.3 ± 5.9	29.2	23.9, 33.2
Abduction (degrees)	Walk	BW	-4.4 ± 3.4	-4	-6.5, -1.4
		+25%BW	-4.7 ± 3.4	-4.9	-6.8, -2.6
		+45%BW	-4.8 ± 3.6	-5	-6.9, -2.8
	Force March	BW	-4.4, 3.6	-4.1	-6.8, -1.4
		+25%BW	-4.5 ± 3.6	-3.8	-6.8, 1.6
		+45%BW	-4.3 ± 3.7	-3.3	-6.6, 1.8
	Run	BW	-5.0 ± 3.7	-4.3	-7.5, -2.0
		+25%BW	4.8 ± 3.6	-3.6	-7.2, -1.9
		+45%BW	-4.7 ± 3.6	-3.4	-6.8, -1.6
Internal Rotation (degrees)	Walk	BW	4.0 ± 6.4	5.3	1.6, 7.3
		+25%BW	4.0 ± 6.7	4.4	1.4, 8.5
		+45%BW	3.5 ± 7.0	5.8	-1.1, 7.7
	Force March	BW	4.9 ± 6.2	4.8	2.2, 6.9
		+25%BW	4.9 ± 5.8	4.8	3.5, 8.3
		+45%BW	7.2 ± 6.0	6.5	4.9, 9.4
	Run	BW	6.8 ± 6.2	6.9	4.4, 10.1
		+25%BW	7.4 ± 6.1	4.9	4.2, 12.1
		+45%BW	9.2 ± 6.3	7.5	5.6, 14.0

Table 5 (continued)

Medial Translation (mm)	**Walk**	**BW**	3.4 ± 1.3	3.4	2.5, 4.5
		+25%BW	3.2 ± 1.5	3.2	1.8, 4.5
		+45%BW	3.3 ± 1.5	3.4	2.0, 4.8
	Force March	**BW**	4.1 ± 1.5	4.4	3.4, 5.2
		+25%BW	3.7 ± 1.3	3.9	3.0, 4.7
		+45%BW	3.6 ± 1.3	3.7	2.4, 4.8
	Run	**BW**	3.9 ± 1.5	3.9	2.6, 5.0
		+25%BW	3.8 ± 1.5	4.3	2.6, 4.8
		+45%BW	4.0 ± 1.6	3.9	2.6, 5.0
Proximal Translation (mm)	**Walk**	**BW**	-22.3 ± 2.3	-23.5	-23.9, -21.1
		+25%BW	-21.9 ± 2.1	-22.4	-23.4, -20.0
		+45%BW	-22.1 ± 2.1	-22.3	-23.8, -20.4
	Force March	**BW**	-21.5 ± 2.2	-21.8	-23.1, -20.1
		+25%BW	-21.5 ± 2.2	-22.2	-22.8, -21.1
		+45%BW	-21.1 ± 2.4	-21.9	-22.8, -18.4
	Run	**BW**	-20.4 ± 2.2	-21.2	-21.7, -18.8
		+25%BW	-20.3 ± 2.2	-20.5	-22.1, '18.3
		+45%BW	-20.3 ± 2.2	-20.6	-22.2, -18.1
Anterior Translation (mm)	**Walk**	**BW**	9.7 ± 3.8	10.1	8.4, 11.1
		+25%BW	10.1 ± 4.3	10.1	8.8, 11.6
		+45%BW	9.8 ± 3.7	9.7	8.2, 11.7
	Force March	**BW**	9.5 ± 4.0	9.3	8.2, 11.4
		+25%BW	9.7 ± 3.8	9.5	8.2, 11.2
		+45%BW	9.9 ± 4.0	10.4	7.9, 11.5
	Run	**BW**	9.3 ± 4.4	9.6	6.3, 11.2
		+25%BW	10.0 ± 3.8	9.9	9.2, 11.5
		+45%BW	10.3 ± 4.1	10.7	8.7, 12.2

There was no significant interaction between load and locomotion, in their effect on flexion ($F_{4,44}$ = 1.341, p = 0.270, η_p^2 = 0.109). There was a significant main effect of locomotion on flexion averaged across levels of load ($F_{1.391,15.297}$ = 32.922, p = <0.001, η_p^2 =0.750) (Figure 7). Flexion was significantly lower at the walking category of locomotion (mean = 17.7˚, SE = 6.3˚) than the running category of locomotion (mean = 27.6˚, SE = 6.7˚), averaged across averaged across levels of load (p <0.001). Flexion was significantly lower at the forced marching category of locomotion (mean = 19.5˚, SE = 6.3˚) than the running category of locomotion (mean = 27.6˚, SE = 6.7˚), averaged across averaged across levels of load (p <0.001). There was no significant main effect of load on flexion, averaged across locomotion.

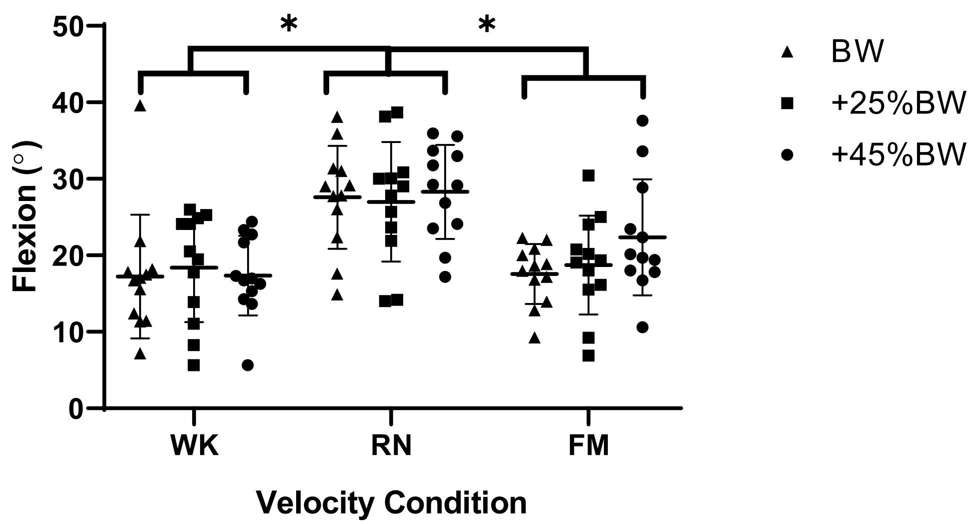

Figure 7: Average flexion at 30% right leg support.

Flexion angle (˚) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

4.1.2.5 Joint Contact

Medial Compartment Minimum Gap at Right Heel Strike

Average medial and lateral compartment minimum gap data for right heel strike is presented in Table 6.

Table 6: Average minimum tibiofemoral gap at right heel strike (mm) (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Medial Plateau	Walk	BW	2.4 ± 0.6	2.4	1.9, 3.0
		+25%BW	2.4 ± 1.1	2.7	1.8, 3.3
		+45%BW	2.3 ± 1.0	2.3	1.7, 2.7
	Force March	BW	1.6 ± 1.2	1.4	0.7, 2.5
		+25%BW	1.8 ± 0.9	1.5	1.0, 2.1
		+45%BW	2.3 ± 0.9	2.3	1.9, 2.6
	Run	BW	3.1 ± 0.8	3.0	2.7, 3.2
		+25%BW	2.8 ± 1.1	3.0	2.1, 3.4
		+45%BW	2.7 ± 1.1	3.0	2.5, 3.2
Lateral Plateau	Walk	BW	3.5 ± 0.8	3.4	2.7, 3.8
		+25%BW	3.3 ± 1.2	3.7	3.2, 4.0
		+45%BW	3.2 ± 1.1	2.9	2.5, 4.0
	Force March	BW	2.5 ± 1.3	2.3	1.9, 3.2
		+25%BW	2.7 ± 0.8	2.7	2.2, 3.2
		+45%BW	3.4 ± 1.2	3.4	2.6, 3.9
	Run	BW	3.9 ± 1.2	3.7	3.4, 4.6
		+25%BW	4.0 ± 0.9	3.8	3.4, 4.6
		+45%BW	3.8 ± 1.0	3.8	3.3, 4.3

There was no significant interaction between load and locomotion in their effect on medial plateau minimum gap ($F_{2.440,26.844} = 1.944$, $p = 0.155$, $\eta_p^2 = 0.150$). There was a significant main effect of locomotion on medial compartment minimum gap averaged across levels of load ($F_{2,22} = 88.255$, $p = 0.002$, $\eta_p^2 = 0.429$) (Figure 8). Medial compartment minimum gap was significantly lower at the forced march category of locomotion (mean = 1.8 mm, SE = 1.1 mm) than the running category of locomotion (mean = 2.9 mm, SE = 1.0 mm), averaged across levels of load (p <0.001). There was no significant main effect of load on medial compartment minimum gap averaged across load.

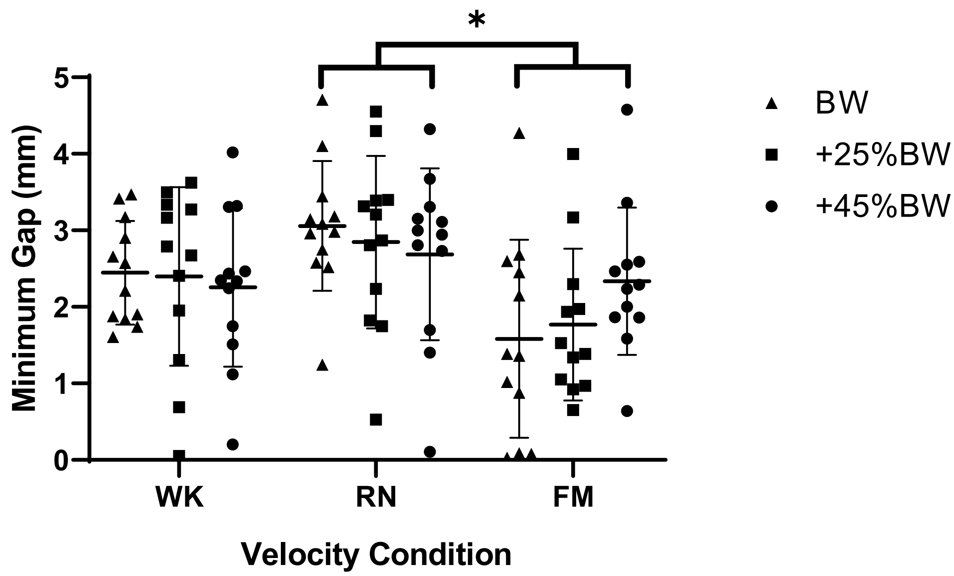

Figure 8: Average medial compartment minimum gap at right heel strike.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Lateral Compartment Minimum Gap at Right Heel Strike

There was no significant interaction between load and locomotion, in their effect on lateral compartment minimum gap ($F_{2.406,26.464} = 1.730$, $p = 0.161$, $\eta_p^2 = 0.136$). There was a significant main effect of locomotion on lateral plateau minimum gap, averaged across levels of load ($F_{2,22} = 9.314$, $p = 0.001$, $\eta_p^2 = 0.458$) (Figure 9). Lateral compartment minimum gap was significantly lower at the forced marching category of locomotion (mean = 2.9 mm, SE = 1.2 mm) than the running category of locomotion (mean = 3.9 mm, SE = 1.0 mm), averaged across levels of load (p <0.001). There was no significant main effect of load, averaged across locomotion.

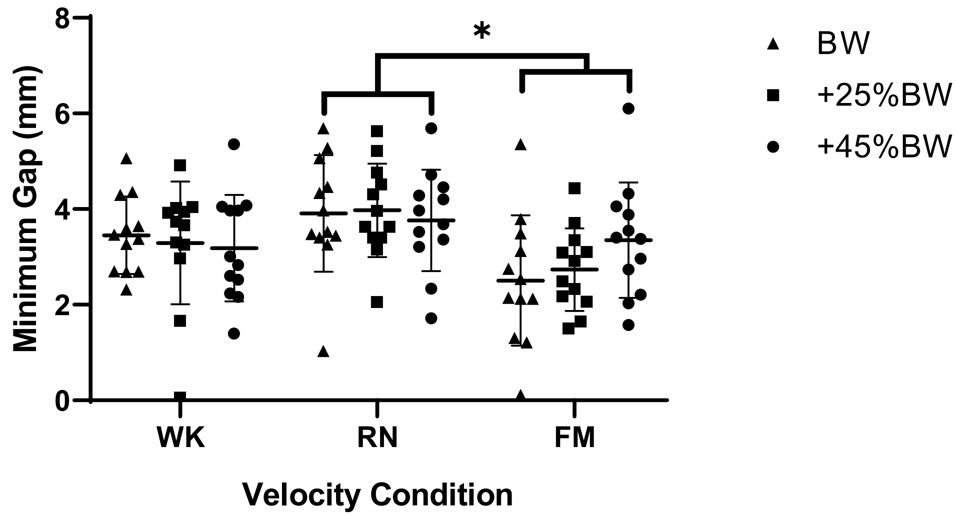

Figure 9: Average lateral compartment minimum gap at right heel strike.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Medial Compartment Minimum Gap at 10% Right Leg Support

Average medial and lateral compartment minimum gap data at 10% right leg support is presented in Table 7.

Table 7: Average minimum tibiofemoral gap at 10% right leg support (mm) (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Medial Plateau	**Walk**	**BW**	2.7 ± 0.9	2.6	2.0, 3.2
		+25%BW	3.1 ± 0.7	2.9	2.7, 3.4
		+45%BW	2.7 ± 0.6	2.6	2.3, 2.9
	Force March	**BW**	2.6 ± 1.1	2.9	2.1, 3.2
		+25%BW	3.0 ± 1.0	2.9	2.7, 3.8
		+45%BW	2.8 ± 0.9	2.7	2.2, 3.0
	Run	**BW**	2.9 ± 0.7	2.9	2.5, 3.1
		+25%BW	3.1 ± 0.7	3.0	2.7, 3.4
		+45%BW	3.1 ± 0.7	2.8	2.6, 3.7
Lateral Plateau	**Walk**	**BW**	3.6 ± 0.9	3.5	2.9, 4.1
		+25%BW	4.0 ± 0.7	3.9	3.6, 4.2
		+45%BW	3.6 ± 0.8	3.6	3.0, 4.1
	Force March	**BW**	3.5 ± 1.1	3.6	3.3, 4.2
		+25%BW	4.0 ± 0.9	4.1	3.3, 4.5
		+45%BW	3.7 ± 0.9	3.5	3.3,3.9
	Run	**BW**	3.6 ± 1.0	3.7	3.2, 4.3
		+25%BW	4.0 ± 0.7	3.9	3.5, 4.7
		+45%BW	4.1 ± 0.6	4.1	3.7, 4.5

There was no significant interaction between load and locomotion, in their effect on medial compartment minimum gap ($F_{4,44} = 0.372$, $p = 0.827$, $\eta_p^2 = 0.033$). There was a significant main effect of load on medial compartment minimum gap, averaged across levels of locomotion ($F_{2,22} = 3.658$, $p = 0.043$, $\eta_p^2 = 0.250$) (Figure 10). There were no significant pairwise comparisons. There was no significant main effect of locomotion on medial compartment minimum gap, averaged across load.

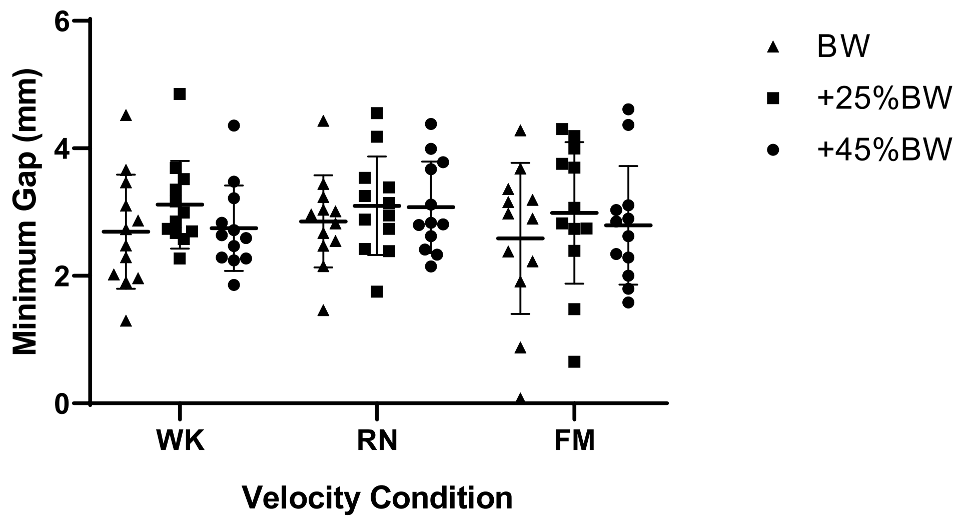

Figure 10: Average medial compartment minimum gap at 10% right leg support.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Lateral Compartment Minimum Gap at 10% Right Leg Support

There was no significant interaction between load and locomotion, in their effect on lateral compartment minimum gap ($F_{4,44} = 0.858$, $p = 0.497$, $\eta_p^2 = 0.072$). There was a significant main effect of load on lateral compartment minimum gap, averaged across levels of locomotion ($F_{2,2} = 3.896$, $p = 0.036$, $\eta_p^2 = 0.262$) (Figure 11). There were no significant pairwise comparisons. There was no significant main effect of locomotion on lateral compartment minimum gap, averaged across load.

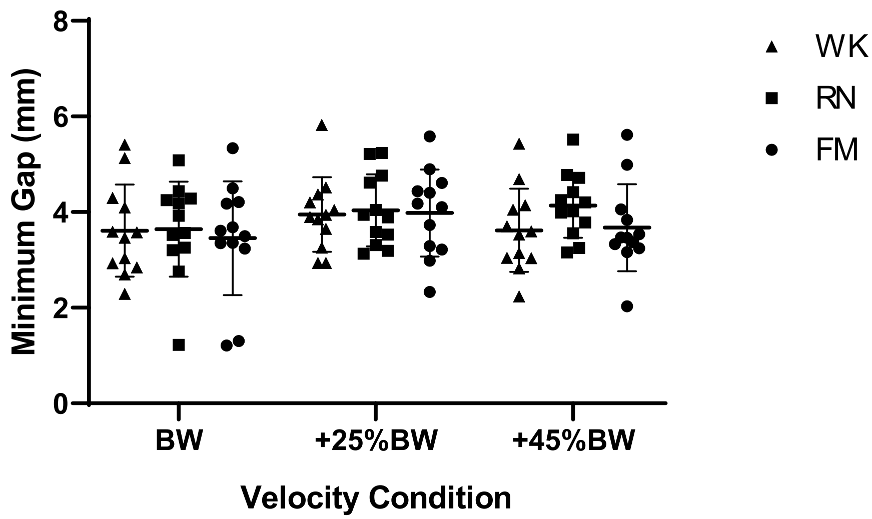

Figure 11: Average lateral compartment minimum gap at 10% right leg support.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Medial Compartment Minimum Gap at 20% Right Leg Support

Average medial compartment minimum gap at 20% right leg support is presented in Table 8.

Table 8: Average minimum tibiofemoral gap at 20% right leg support (mm) (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Medial Plateau	**Walk**	**BW**	2.8 ± 0.7	2.7	2.4, 3.1
		+25%BW	2.9 ± 0.7	2.9	2.2, 3.2
		+45%BW	2.6 ± 0.6	2.4	2.3, 2.6
	Force March	**BW**	2.8 ± 0.7	2.7	2.4, 3.2
		+25%BW	2.7 ± 0.7	2.4	2.2, 3.0
		+45%BW	2.5 ± 0.6	2.5	1.9, 2.7
	Run	**BW**	2.6 ± 0.9	2.5	1.8 2.9
		+25%BW	2.6 ± 0.6	2.5	2.0, 3.0
		+45%BW	2.5 ± 0.6	2.4	2.1, 2.9
Lateral Plateau	**Walk**	**BW**	3.6 ± 0.7	3.4	3.0, 4.2
		+25%BW	3.7 ± 0.6	3.7	3.3, 4.1
		+45%BW	3.6 ± 0.7	3.4	3.1, 3.9
	Force March	**BW**	3.6 ± 0.8	3.5	3.0, 4.0
		+25%BW	3.7 ± 0.6	3.7	3.3, 4.1
		+45%BW	3.6 ± 0.7	3.4	3.1, 3.9
	Run	**BW**	3.5 ± 0.7	3.6	2.9, 3.8
		+25%BW	3.6 ± 0.7	3.5	3.0, 4.1
		+45%BW	3.6 ± 0.7	3.6	3.1, 3.8

There was no significant interaction between load and locomotion in their effect on medial compartment minimum gap ($F_{1.996,21.960} = 0.573$, $p = 0.572$, $\eta_p^2 = 0.050$. There was a significant main effect of load on medial compartment minimum gap, averaged across levels of locomotion ($F_{2,22} = 4.141$, $p = 0.030$, $\eta_p^2 = 0.273$) (Figure 12). Medial compartment minimum gap was significantly lower at the +45%BW loading condition (mean = 2.5 mm, SE = 0.6 mm) than the +25%BW loading condition (mean = 2.7 mm, SE = 0.8 mm), averaged across levels of locomotion ($p < 0.001$). There was no significant main effect of locomotion on medial compartment minimum gap, averaged across load.

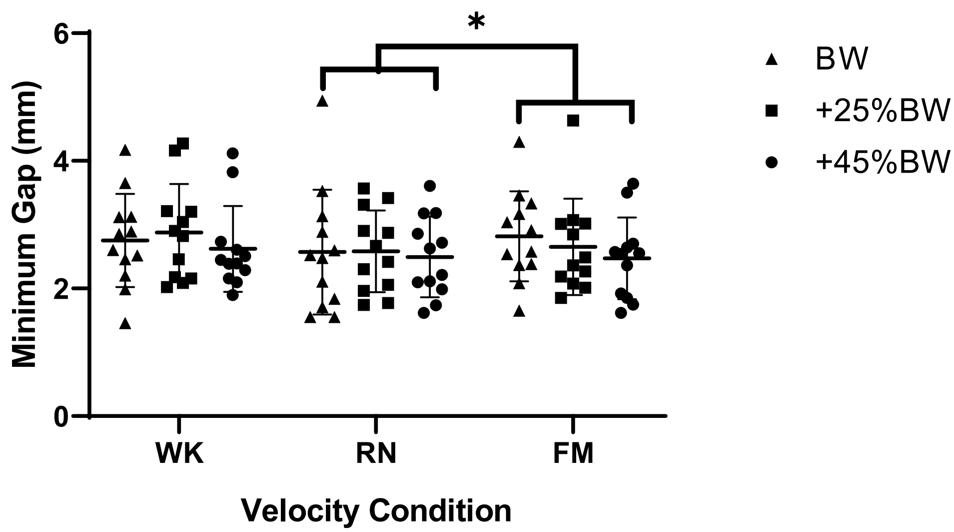

Figure 12: Average medial compartment minimum gap at 20% right leg support.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW

Lateral Compartment Minimum Gap at 20% Right Leg Support

There was no significant interaction between load and locomotion, in their effect on lateral compartment minimum gap ($F_{4,44}$ = 0.373, p = 0.827, η_p^2 = 0.033). There was no significant main effect of load on lateral compartment minimum gap, averaged across locomotion. There was no significant main effect of locomotion on lateral compartment minimum gap, averaged across load (Figure 13).

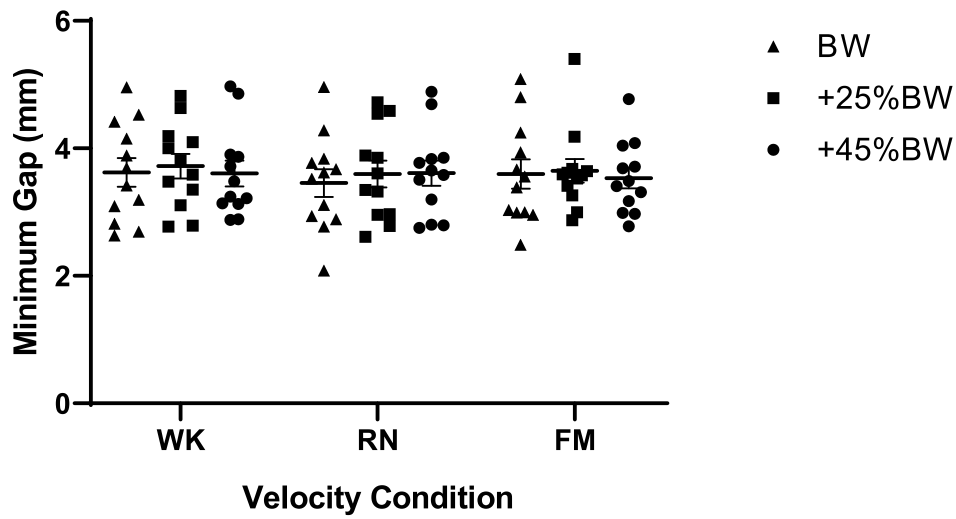

Figure 13: Average lateral compartment minimum gap at 20% right leg support.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Medial Compartment Minimum Gap at 30% Right Leg Support

Average medial compartment minimum gap data at 30% right leg support is presented in Table 9.

Table 9: Average minimum tibiofemoral gap at 30% right leg support (mm) (N=12).

		Load (%BW)	Mean ± SD	Median	IQR (Q1, Q3)
Medial Plateau	Walk	BW	2.6 ± 0.6	2.5	2.2, 3.0
		+25%BW	2.5 ± 0.6	2.3	2.2, 2.5
		+45%BW	2.5 ± 0.5	2.3	2.1, 2.6
	Force March	BW	2.0 ± 0.8	1.8	1.6, 2.6
		+25%BW	2.2 ± 0.7	1.9	1.5, 2.6
		+45%BW	2.1 ± 0.5	1.9	1.8, 2.3
	Run	BW	2.1 ± 0.7	2.1	1.7, 2.4
		+25%BW	2.1 ± 0.6	2.1	1.8, 2.3
		+45%BW	2.2 ± 0.9	2.0	1.8, 2.5
Lateral Plateau	Walk	BW	4.0 ± 0.7	3.9	3.6, 4.8
		+25%BW	3.7 ± 0.7	3.4	3.2, 4.2
		+45%BW	3.8 ± 0.6	3.4	3.2, 4.3
	Force March	BW	3.3 ± 1.0	3.5	2.7, 3.8
		+25%BW	3.5 ± 0.8	3.3	2.9, 4.0
		+45%BW	3.6 ± 0.6	3.5	3.2, 4.1
	Run	BW	3.4 ± 0.7	3.2	2.8, 3.8
		+25%BW	3.4 ± 0.7	3.2	2.9, 3.8
		+45%BW	3.6 ± 0.9	3.1	2.9, 3.9

There was no significant interaction between locomotion and load in their effect on medial compartment minimum gap ($F_{4,44} = 0.950$, $p = 0.445$, $\eta_p^2 = 0.079$). There was a significant main effect of locomotion on medial plateau minimum gap, averaged across levels of load ($F_{2,22} = 11.672$, $p = 0.000$, $\eta_p^2 = 0.515$) (Figure 14). The level of medial compartment minimum gap was significantly lower at the forced march category of locomotion (mean = 2.1 mm, SE = 0.7 mm) than the walking category of locomotion (mean = 2.5 mm, SE = 0.6 mm), averaged across levels of load ($p < 0.001$). Medial compartment minimum gap was significantly lower at the running category of locomotion (mean = 2.1 mm, SE = 0.7) than the walking category of locomotion (mean = 2.5 mm, SE = 0.6 mm), averaged across levels of load (p

<0.001). There was no significant main effect of load on medial compartment minimum gap, averaged across locomotion.

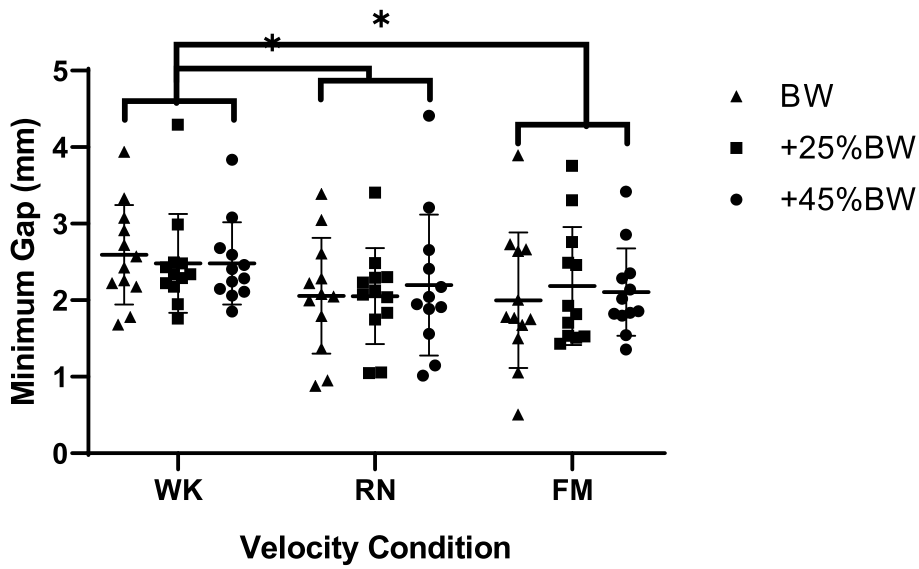

Figure 14: Average medial compartment minimum gap at 30% right leg support.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Lateral Compartment Minimum Gap at 30% Right Leg Support

There was a significant interaction between load and locomotion, in their effect on lateral plateau minimum gap ($F_{4,44} = 2.907$, $p = 0.032$, $\eta_p^2 = 0.209$) (Figure 15). Simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at the bodyweight level of load ($F_{2,22} = 8.354$, $p = 0.002$, $\eta_p^2 = 0.432$). There was a significant main effect of locomotion at the +25%BW level of load ($F_{2,22} = 5.005$, $p = 0.016$, $\eta_p^2 = 0.313$).

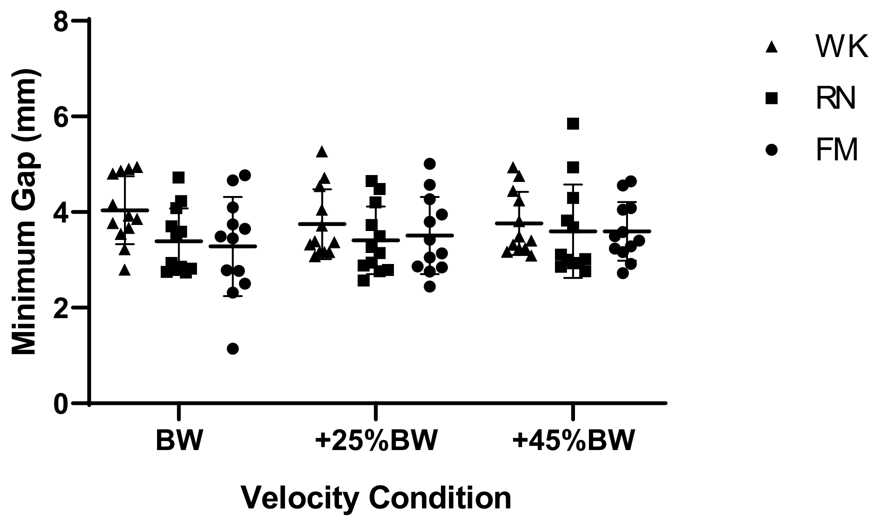

Figure 15: Average lateral compartment minimum gap at 30% right leg support.

Minimum gap (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Contact Path Length

Average contact path length for the medial and lateral plateau and contact path length normalized to plateau height and width are presented in Table 10.

Table 10: Raw and normalized contact path length (mm) over 0-30% right leg support. (N=12)

			Raw Length (mm)			**Length Normalized to Plateau Width, Height (mm)**		
		Load (%BW)	**Mean ± SD**	**Median**	**IQR (Q1, Q3)**	**Mean ± SD**	**Median**	**IQR (Q1, Q3)**
Medial Plateau	**Walk**	**BW**	9.9 ± 4.7	8.7	6.8, 11.7	0.3 ± 0.2	0.3	0.2, 0.4
		+25%BW	11.9 ± 4.9	12.3	9.1, 13.0	0.4 ± 0.2	0.4	0.2, 0.4
		+45%BW	10.1 ± 4.2	8.8	7.9, 10.9	0.3 ± 0.1	0.3	0.2, 0.4
	Force March	**BW**	11.4 ± 4.3	10.8	9.0, 12.5	0.3 ± 0.1	0.3	0.3, 0.4
		+25%BW	11.5 ± 4.4	11.2	7.4, 14.2	0.4 ±0.1	0.4	0.2, 0.4
		+45%BW	12.6 ± 3.8	12.4	8.7, 16.8	0.4 ± 0.1	0.4	0.3, 0.5
	Run	**BW**	9.9 ± 2.4	10.2	8.0, 11.9	0.3 ± 0.1	0.3	0.2, 0.3
		+25%BW	9.3 ± 2.0	9.3	8.1, 11.3	0.3 ± 0.1	0.3	0.2, 0.3
		+45%BW	10.2 ± 2.3	10.1	8.4, 12.0	0.3 ± 0.1	0.3	0.3, 0.4
Lateral Plateau	**Walk**	**BW**	7.9 ± 2.2	7.2	6.0, 9.3	0.3 ± 0.1	0.3	0.2, 0.3
		+25%BW	9.2 ± 3.3	9.1	7.3, 10.1	0.3 ± 0.1	0.3	0.2, 0.3
		+45%BW	8.5 ± 2.7	8.3	6.3, 9.9	0.3 ± 0.1	0.3	0.2, 0.4
	Force March	**BW**	8.8 ± 1.8	8.3	7.2, 9.8	0.3 ± 0.1	0.3	0.2, 0.3
		+25%BW	9.2 ± 1.6	8.8	7.8, 11.1	0.3 ± 0.1	0.3	0.3, 0.4
		+45%BW	10.0 ± 1.7	9.8	8.3, 11.4	0.3 ± 0.1	0.3	0.2, 0.4
	Run	**BW**	7.5 ± 1.9	7.4	6.6, 8.2	0.2 ± 0.1	0.2	0.2, 0.3
		+25%BW	8.5 ± 1.1	8.4	7.9, 9.2	0.3 ± 0.0	0.2	0.2, 0.3
		+45%BW	9.0 ± 1.9	9.1	8.1, 10.6	0.3 ± 0.1	0.3	0.2, 0.3

Medial Plateau Normalized Contact Path Length

There was no significant interaction between load and locomotion, in their effect on medial plateau contact path length ($F_{4,44} = 1.313$, $p = 0.280$, $\eta_p^2 = 0.107$). There was a significant main effect of locomotion on medial plateau contact path length, averaged across levels of load ($F_{2,22} = 3.5427$, $p = 0.046$, $\eta_p^2 = 0.244$) (Figure 16). There were no significant pairwise comparisons. There was no significant main effect of load on medial plateau contact path length, averaged across locomotion.

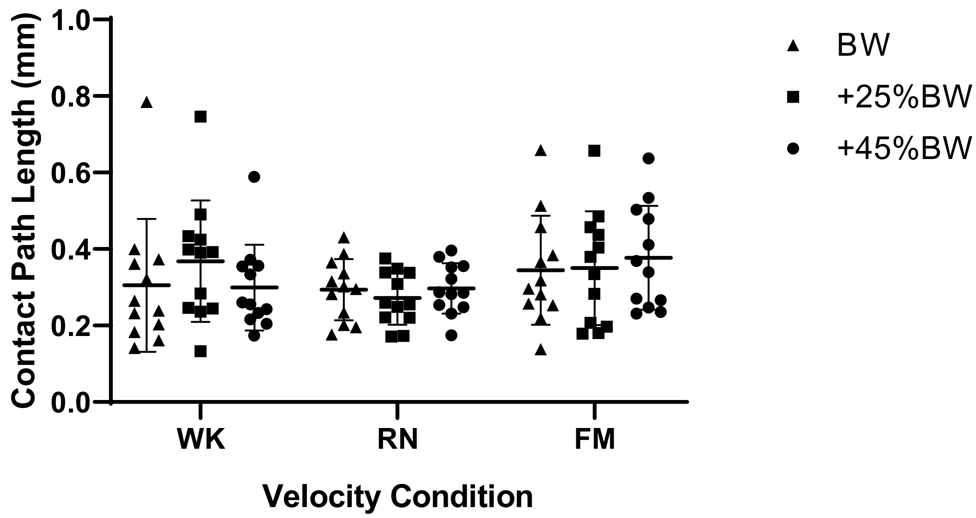

Figure 16: Average medial compartment normalized contact path length.

Contact path length (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Lateral Plateau Normalized Contact Path Length

There was no significant interaction between load and locomotion, in their effect on lateral plateau contact path length ($F_{4,44} = 0.636$, $p = 0.639$, $\eta_p^2 = 0.055$). There was a significant main effect of locomotion on lateral plateau contact path length, averaged across levels of load ($F_{2,22} = 4.412$, $p = 0.024$, $\eta_p^2 = 0.286$) (Figure 17). Lateral plateau contact path length was significantly lower at the running category of locomotion (mean = 0.3 mm, SE = 0.1 mm) than the forced march category of locomotion (mean = 0.3 mm, SE = 0.1 mm), averaged across levels of load ($p <0.001$). There was no significant main effect of load on lateral plateau contact path length, averaged across levels of locomotion.

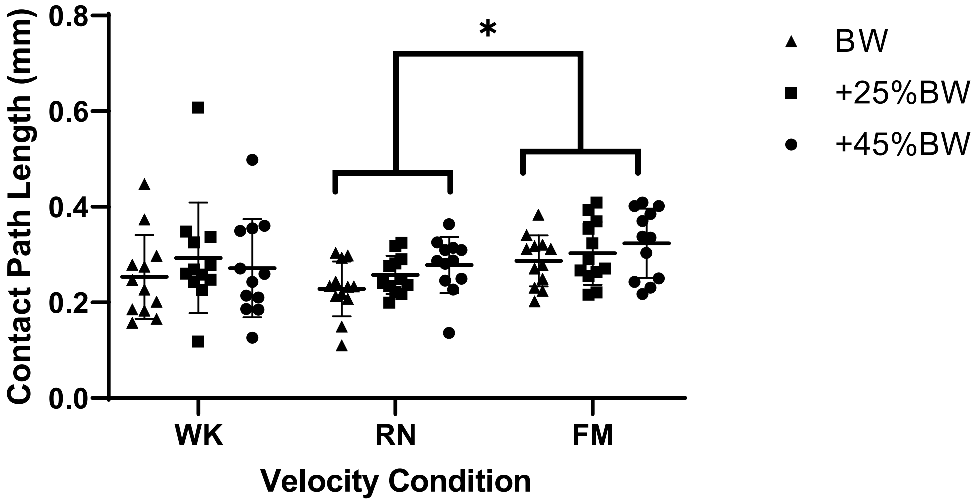

Figure 17: Average lateral compartment normalized contact path length.

Contact path length (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

5.0 Discussion

The current study investigated the effects of load magnitude and marching velocity on tibiofemoral kinematics in recruit-aged women. The addition of load has the potential to disrupt normal gait patterns while running and walking, so addressing increased relative loads as well as high velocity marching (forced marching) common to military training could provide insight into stable gait patterns and prevention of injury and long-term joint degeneration.

Fourteen moderately physically active female subjects were enrolled in the study. One subject neglected to schedule a data collection visit following the orientation session and a second subject was only able to perform one data collection session due to equipment malfunction. Overall, twelve subjects were included in data analysis. Subjects were asked to perform locomotion trials on an instrumented treadmill where they walked, ran, or marched at a high velocity while unloaded, loaded with additional 25% bodyweight, and loaded with additional 45% bodyweight. Tibiofemoral arthrokinematics during loaded stance phase were quantified using biplane radiography and custom model-based tracking software to calculate *in-vivo* bone motion. Data were interpolated to percent right leg support (right heel strike to right toe off) and tibiofemoral rotations, translations, contact path, and joint space were analyzed from right heel strike (0%) to 30% right leg support in intervals of 10%. A two-way repeated measures ANOVA was used to analyze the interactive effects of load carriage magnitude and gait velocity on these biomechanical variables.

Three specific aims were addressed through this analysis. The first specific aim was to determine the effects of increased load on tibiofemoral arthrokinematics (unloaded, loaded with additional 25% bodyweight, loaded with additional 45% bodyweight). It was hypothesized that increased load would be associated with increased flexion, decreased dynamic joint space, and increased contact path length. This hypothesis was not supported by the results. The second specific aim was to determine the effects of different locomotion velocities on tibiofemoral arthrokinematics (walking, running, and forced march at a high velocity). It was hypothesized that increased velocity would be associated with increased flexion, with the most identifiable change seen in the forced marching condition, and that contact path would shift posteriorly with increased velocity. This hypothesis was partially supported by the results. The final specific aim was to identify the interactive effects of load carriage and marching velocity on knee arthrokinematics. It was hypothesized that there would be a significant interaction of load and locomotion velocity for flexion, joint space, and contact path. This hypothesis was partially supported by the results. The effects of load and locomotion velocity and limitations of the current study will be addressed in the following sections.

5.1 The Effects of Load Magnitude on Tibiofemoral Arthrokinematics

Overall, increased load had no significant effects on knee flexion, which conflicts with previously published research and does not support our hypothesis of increased knee flexion with increasing relative load. Locomotion velocity was found to be more influential when analyzing knee flexion at different percentages of the right leg support phase.

Increasing load resulted in increased average tibiofemoral flexion for all measured time points, but the results were not statistically significant. The BW load condition had the lowest amount of flexion for all locomotion categories (11.6°, 8.6°, 13.9°, and 20.8° at 0-30% of right leg support), indicating increased flexion as the limb approached midstance. The +25%BW condition showed a similar trend (15.0°, 9.8°, 15.0°, and 21.3° at 0-30% of right leg support), as did the +45%BW condition (15.1°, 10.5°, 15.0°, 22.7°). Flexion at 10% of right leg support was consistently lowest among all loading groups, indicating extension following right heel strike before moving further into support phase.

Previous research has found that flexion at heel strike significantly increased with additional loading.(Attwells et al., 2006; Birrell & Haslam, 2009) However, differences in data collection and analysis procedures may contribute to this discrepancy, as well as the participant demographics and loads carried. The average load used in this study was 14.3 ± 2.0 kg for the +25%BW load condition and 25.6 ±3.5 kg for the +45%BW load condition. Loads of approximately 35% of a soldier's bodyweight are more commonly used as fighting loads, with heavier approach march loads and combat loads approaching 57-70% of the *average male* soldier's bodyweight.(Schuh-Renner et al., 2017) For the purpose of this analysis, loads were relative to participant bodyweight for several reasons. First, analyzing an individual's response to a relative load allowed for comparison of tibiofemoral arthrokinematics across individuals of different weights and sizes. Additionally, the relatively light +45%BW load is comparable to what new recruits may use to begin their load carriage training and would be safe to use in our study population of inexperienced carriers. The use of the weighted vest as a carrying mechanism helped to limit the confounding center of mass shift that occurs when carrying a load that is unequally distributed about the participant, such as a backpack load, but is not representative of the load distribution found from military load carriage systems.

Increased load significantly decreased minimum gap distance for the medial and lateral compartments, but only for certain portions of the analyzed support phase. An increased load resulted in significantly decreased medial and lateral compartment minimum gap at 10% of right heel strike, but no significant pairwise comparisons were identified. Average medial compartment minimum gap at this time point was 2.7 mm, 3.1 mm, and 2.9 mm for BW, +25%BW, and +45% BW, respectively. Average lateral compartment minimum gap at this time point was 3.6 mm, 4.0 mm, and 3.9 mm for BW, +25%BW, and +45%BW, respectively. Medial compartment minimum gap at 20% right leg support was significantly lower during the +45%BW loading condition compared to +25%BW (average difference: 0.1 mm). This effect was not seen in the lateral compartment at the same time point. The significant main effect of load at 10-20% right leg support may indicate that this period of early stance phase following initial foot contact may elicit the greatest change in joint space. This period of "load acceptance", where the carrier absorbs the resulting forces from contact with the ground via their musculature and biomechanical compensations could provide insight into strategies of load mitigation and dissipation through training protocols which target lower limb musculature and coordination. Further analysis is needed to explore this concept further.

The lack of statistically significant differences in compartmental minimum gap between +25%BW and +45%BW bodyweight load conditions for most of the studied time points could suggest that the studied loads are not large enough to affect detectable changes in tibiofemoral arthrokinematics, or potentially that the detectable changes were within the measurement error of the imaging and analysis system. Additionally, the study population may be physically fit enough to handle the relatively light loads used in this study. Since all participants were moderately physically active it is possible that they had sufficient musculature and neuromuscular control to dampen the effects of loading. Incorporating higher loads or less fit participants may elicit identifiable changes in tibiofemoral biomechanics with the addition of load and changing velocity conditions.

One potential factor that could influence compartmental loading that was not analyzed during this study is static knee alignment. The results of this study indicate that the compartments of the knee may be affected differently by increased loading or a change in locomotion strategy. If there is an increased stress on the lateral compartment during stance phase, this loading could potentially be linked to an increased risk of lateral knee osteoarthritis, particularly when combined with the increase in impact during a running gait as opposed to a walking gait. Previous studies have demonstrated that individuals with greater knee valgus have higher incidences for lateral compartment knee osteoarthritis (KOA) when compared to people with a normal or varus knee alignment, with women experiencing greater rates of lateral KOA compared to men.(Barrios et al., 2016; Brouwer et al., 2007) For this reason, classifying participants by degree of knee valgus is required to determine if knee valgus without additional traumatic injury to the knee results in altered gait biomechanics contributing to an increased risk in lateral KOA over an individual's life, particularly in individuals such as military members who are held to high standards of performance throughout their careers.(Hoch & Weinhandl, 2017) The fact that medial compartment gap was not consistently affected by the increase in load supports current research.

Similar to the results for medial and lateral joint space, load had no effect on medial or lateral contact path length. An increase in load resulted in normalized medial contact path lengths of 0.32, 0.33, and 0.33 for BW, +25%BW, and +45%BW, showing no statistically significant effect of additional load on relative motion of the femur and tibia. In the lateral compartment, an increase in load resulted in normalized contact path lengths of 0.26, 0.28, and

0.29 for BW, +25%BW, and +45%B, respectively. No significant differences based on load magnitude were detected. Since no statistically different flexion angles were found based on an increase in carried load, it follows that the contact path lengths would also be largely unaffected in a population of healthy knees (i.e. no excessive translation due to an underlying pathology at the joint).

One possible contributing factor to the lack of significant effects of load and velocity on compartmental gap is gait variability. Gait variability between individuals results in differing knee flexion angles when analyzing discrete time points during the support phase. If possible, analyzing the effects of load and locomotion on compartmental gap at specific flexion angles rather than time points could provide a more comprehensive view of how joint space changes with tibiofemoral kinematics. To expand upon this investigation, cartilage models from magnetic resonance imaging could be used as another way of identifying compression occurring during this portion of gait. Further investigation into the mechanism of cartilage contact while carrying heavy loads is warranted to understand the potential long-term degenerative effects of military-relevant load carriage in healthy young women.

5.2 The Effects of Gait Velocity on Tibiofemoral Arthrokinematics

Increased velocity resulted in significant increases in knee flexion across all analyzed time points. When significant differences in flexion were observed, flexion was higher in the forced marching and running conditions when compared to the walking conditions. However, differences in knee flexion angle while walking and forced marching were not statistically significant during 20% right leg support. This partially supports our hypothesis of increased flexion with an increase in locomotion velocity. The transition from a walking to running gait pattern necessitates greater flexion to stabilize the body during heel strike, early stance and load acceptance, and midstance to compensate for the lack of an additional supporting limb, which is seen across most of the measured support phase.

At right heel strike an increase in locomotion velocity resulted in an increase of knee flexion. Flexion angle increased by 7.3° and 7.6° from WK to RN and WK to FM, respectively. The change in locomotion strategy from forced marching to running decreased flexion angle by 0.3°. At 10% right leg support an increase in locomotion velocity resulted in a 9.9° increase of knee flexion from walking to running. At the same time point, knee flexion while running was found to be 6.8° greater than while forced marching, even though the treadmill was at the same speed. At 20% right leg support, an increase in locomotion velocity resulted in an 8.5° increase in knee flexion from walking to running. Knee flexion also increased by 7.5° between forced marching and running (FM<RN) and 1.0° between walking and forced marching (WK<FM) but the results were not significant. At 30% right leg support, an increase in locomotion velocity led to an increase in knee flexion (WK<RN). A change in locomotion strategy from forced marching to running also showed a 1.9° increase in knee flexion (FM<RN).

In reference to joint space, medial and lateral compartment minimum gap decreased during the forced marching when compared to the running at right heel strike. Joint space decreased by 1.0 mm in both the medial and lateral compartments. This shows that, at heel strike, different patterns of locomotion at same locomotion velocity may affect the knee cartilage compression differently. Since spatiotemporal parameters of gait vary between running and walking these changes may help contribute to differences in loading forced marching and running. Medial compartment minimum gap also decreased by an average of 0.4 mm at 30% right leg support when forced marching compared to walking.

These results are supported by previous research into cartilage contact area during level walking and downhill running in healthy knees, where the medial compartment exhibited greater cartilage contact area than the lateral compartment during both activities.(Akpinar et al., 2019) Decreases in cartilage contact area were also seen in the running trials compared to walking during the beginning portion of the gait cycle.(Akpinar et al., 2019) Although the current study was limited to level walking, it seems as though the medial compartment may be more affected by locomotion velocity than the lateral compartment, yet both are affected by the increased loading that is present during running (i.e. supporting the carrier on a single limb). However, it is again important to note that lack of significant pairwise comparisons at all other analyzed time points indicates a potentially underpowered study or observable changes within the error of the measurement system.

An increase in velocity did not significantly affect the normalized contact path length in the medial compartment; lateral compartment contact path was significantly shorter while running versus forced marching. This change in locomotion strategy resulted in a decrease of 0.05 for lateral normalized contact path length while running compared to walking. Previous research has found significant differences in contact path length during downhill running, but this pattern was not seen with the level running performed during this study.(Gale & Anderst, 2019) Although this pattern was not seen in the medial compartment contact path length data, it follows that an individual who lands with a more flexed knee while running would move through a smaller range of motion than if they had landed with a more extended knee. Flexion was significantly increased at right heel strike for the running and forced marching trials when compared to walking, so it is possible that an increased flexion angle throughout the measured portion of gait leads to a shorter contact path in each compartment.

5.3 The Interactive Effects of Load Magnitude and Gait Velocity

Significant interaction between load carriage magnitude and gait velocity were only found for lateral compartment minimum gap, analyzed at 30% of right leg support. Significant simple main effects were identified at the bodyweight and +25%BW loading conditions. The lack of interactive effects is potentially due to the lack of enough participants to power the study, which should be addressed in future analyses. Additionally, a majority of the identifiable kinematic changes were associated with an increase in velocity rather than load, suggesting that the carrier's familiarity with marching at a high velocity or running could be more influential than the addition of load. Recruiting subjects with different levels of physical activity or load carriage experience may help further identify this relationship, as well as expanding the range of carried load to be more military-relevant (i.e. an absolute load that is carried by all individuals).

5.4 Limitations

The primary limitation of this study is its small sample size. It is possible that the lack of enough subjects to adequately power the study resulted in fewer detectable changes due to load condition. Although the original intent was to analyze 16 subjects to power the interaction of load and velocity, equipment difficulties and attrition contributed to the eventual collection and analysis of 12 subjects. Future work on this topic should consider targeting additional subjects for recruitment to ensure sufficient statistical power to detect changes in tibiofemoral arthrokinematics. This analysis was also limited to one step per movement trial, so collecting and analyzing additional steps may help determine statistical significance as well. An additional limitation is the focus on a single limb during a small portion of the gait cycle. Data collection was limited to a single leg due to restraints from the biplane radiography system, as additional data collection would significantly increase collection time and radiation exposure. Performing data collection using multiple system configurations to image the knee separately during the stance and swing phases of gait would allow for a more complete analysis of tibiofemoral arthrokinematics during different loading and marching conditions. Also, analyzing each limb separately could also help provide a more comprehensive view of carrier stability and identify any differences in the dominant and non-dominant limbs. Although previous studies have addressed the minimal side to side differences in knee kinematics for healthy subjects during walking, verifying that this holds true for loaded, running, and forced marching conditions will provide a more comprehensive dataset regarding knee kinematics in these locomotion conditions. Additionally, analyzing the continuous kinematic data rather than at discrete points may allow for insight into carrier stability and locomotion patterns that are not available through the current analysis, as well as expanding the analysis to include joints such as the hip and ankle.

Subject recruitment characteristics are another consideration when determining an appropriate population to study military load carriage activities. Recruitment was focused on recreationally active females who were not required to have any load carriage experience. Recruiting participants who are familiar with load carriage, whether through recreational activities such as backpacking or through military experience may show observable differences in stability and could speak to how various training methodologies affect the carrier's response to incremental increases in load and velocity. The participant's use of combat boots may manifest similarly, as those with military experience may be more adjusted to the boots and exhibit a more "natural" gait than those who are unfamiliar with military footwear. This may be particularly evident in the more significantly affected phases of initial contact and early midstance when changes in ankle joint angles may be more extreme. Similarly, comparing the effects of additional loading with military-specific loading systems such as backpacks and hip belts in a mixed population of males and females with and without load carriage experience would be a more comprehensive study of how individuals react to identical loading situations.

5.5 Future Research

Future studies should address the effect of load carriage experience and types of training on the effects of military-relevant loads and lower extremity biomechanics. With the population analyzed in this study it was not possible to group subjects based on load carriage experience since load carriage experience was not a criterion for inclusion in the study.

Since few significant interactions between load magnitude and velocity conditions were observed in this very controlled setting, the next step would be to attempt to push the boundaries and elicit conditions that are more similar to activities that warfighters would perform in the field. To make the study more operationally relevant, incorporating an absolute, combat-level load and incorporating different load carriage systems such as waist packs, headgear, and held firearm models may provide an indication of which component of the load carriage process is potentially detrimental to lower extremity function during situations of heavy load carriage.

An in-depth investigation into observable differences in knee kinematics between male and female warfighters is also necessary in order to draw conclusions regarding female-specific injury prevention tactics for military populations. Investigating correlations between anthropometric features such as degree of knee valgus or varus, pelvic Q angle, leg and torso length, foot type, and others could also help identify which characteristics result in biomechanical differences during locomotion. These features could then be used to separate "performers" from "non-performers", that is, those who respond well to load carriage and those who may require additional training and conditioning in order to perform the tasks safely. Addressing potential effects of fatigue during sustained marching activities may provide insight into soldier stability and mobility in the field and establish a safe training threshold for load carriage activities in order to avoid injury.

5.6 Conclusions

The purpose of this analysis was to determine the individual and interactive effects of increased relative load carriage magnitude and gait velocity on tibiofemoral arthrokinematics. Load carriage was most influential when observing joint space in the medial and lateral compartments of the knee, while increased gait velocity had a significant effect on knee flexion and compartment loading at select portions of the early support phase. Limited interaction effects were observed between the increase in load carriage and locomotion velocity.

Appendix

Tibiofemoral Kinematics at Right Heel Strike

Abduction

There was no significant interaction between load and locomotion in their effect on abduction ($F_{4,44} = 1.137$, $p = 0.352$, $\eta_p^2 = 0.094$). There was no significant main effect of load on abduction, averaged across load. There was no significant main effect of locomotion on abduction, averaged across locomotion.

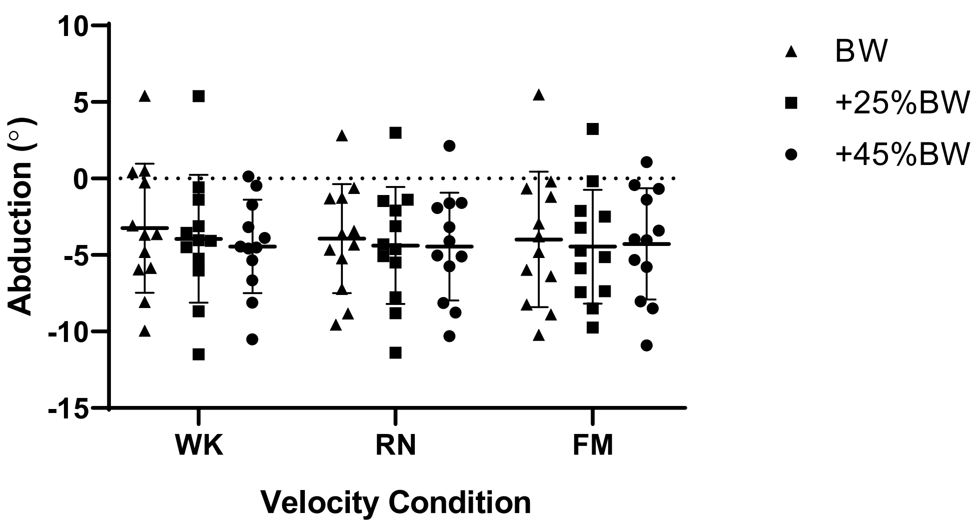

Figure 18: Average tibiofemoral abduction at right heel strike.

Abduction (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Internal Rotation

There was no significant interaction between load and locomotion, in their effect on internal rotation ($F_{4,44} = 1.905$, $p = 0.126$, $\eta_p^2 = 0.148$). There was a significant main effect of locomotion on internal rotation, averaged across levels of load ($F_{2,22} = 4.034$, $p = 0.032$, $\eta_p^2 = 0.268$). There were no significant pairwise comparisons. There was no significant main effect of load on internal rotation, averaged across locomotion.

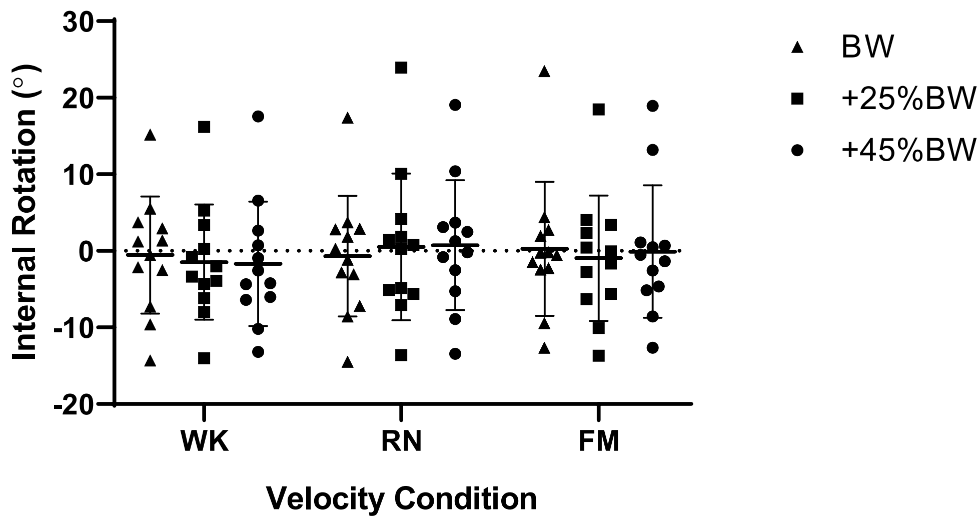

Figure 19: Average tibiofemoral internal rotation at right heel strike.

External rotation (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Medial Translation

There was no significant interaction between and locomotion, in their effect on medial translation ($F_{4,44} = 1.171$, $p = 0.337$, $\eta_p^2 = 0.096$). There was no significant main effect of load on medial translation, averaged across locomotion. There was no significant main effect of locomotion on medial translation, averaged across load.

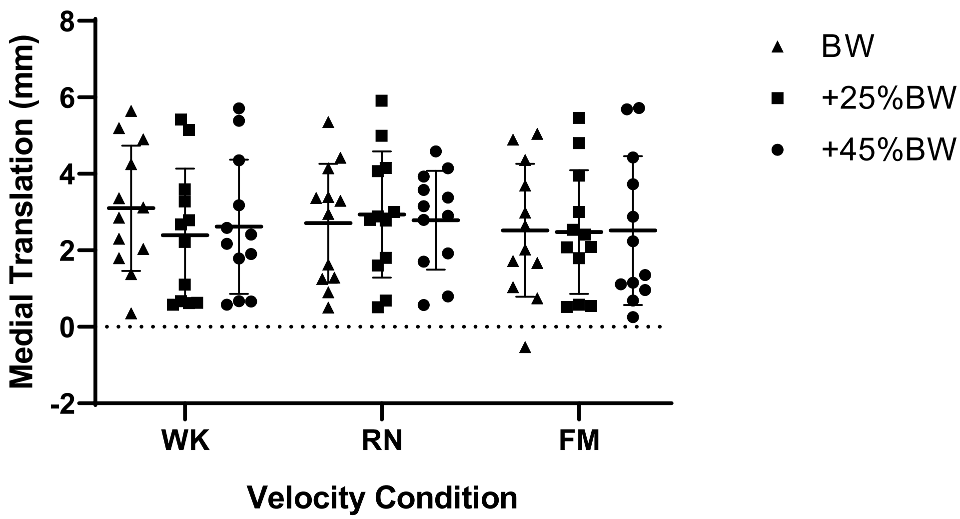

Figure 20: Average tibiofemoral medial translation at right heel strike.

Medial translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Proximal Translation

There was no significant interaction between load and locomotion, in their effect on proximal translation ($F_{4,44}$ = 1.363, p = 0.262, η_p^2 = 0.110). There was a significant main effect of locomotion on proximal translation, averaged across levels of load ($F_{2,22}$ =6.504, p = 0.006, η_p^2 = 0.372). Proximal translation was significantly lower at the walking level of locomotion (-23.6±2.6 mm) than the forced marching level of locomotion (-22.1±2.7mm), averaged across averaged across levels of load (p = 0.013). There was no significant main effect of load on proximal translation, averaged across locomotion.

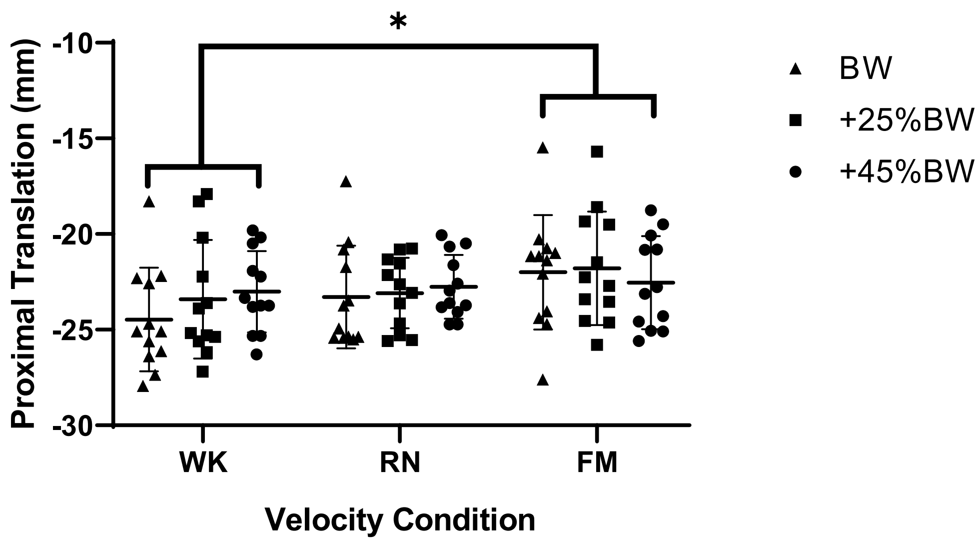

Figure 21: Average tibiofemoral proximal translation at right heel strike.

Proximal translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Anterior Translation

There was no significant interaction between load and locomotion, in their effect on anterior translation ($F_{2.543,27.969}$ = 0.209, p = 0.860, η_p^2 = 0.019). There was a significant main effect of load on anterior translation, averaged across levels of locomotion ($F_{2,22}$ = 4.570, p = 0.022, η_p^2 = 0.293). The level of anterior translation was significantly lower at the +25%BW level of load (4.8±2.7mm) than the bodyweight level of load (5.5±3.1mm), averaged across averaged across levels of locomotion (p = 0.022).

There was a significant main effect of locomotion on anterior translation, averaged across levels of load ($F_{2,22}$ = 6.844, p = 0.005, η_p^2 = 0.384). The level of anterior translation was significantly lower at the running category of locomotion (4.8±2.8mm) than the walking category of locomotion (5.8±3.2mm), averaged across averaged across levels of load (p = 0.026).

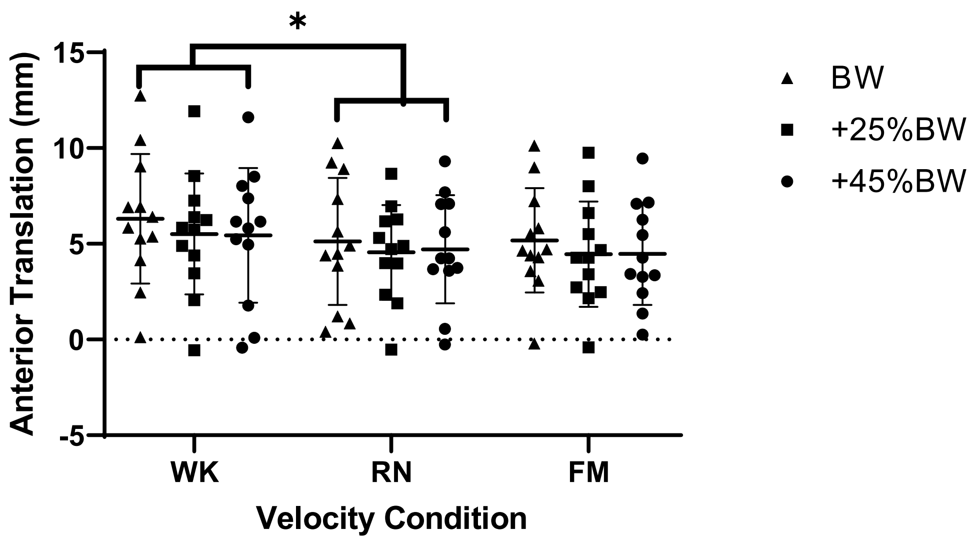

Figure 22: Average tibiofemoral anterior translation at right leg support, pairwise comparisons for load.

Anterior translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

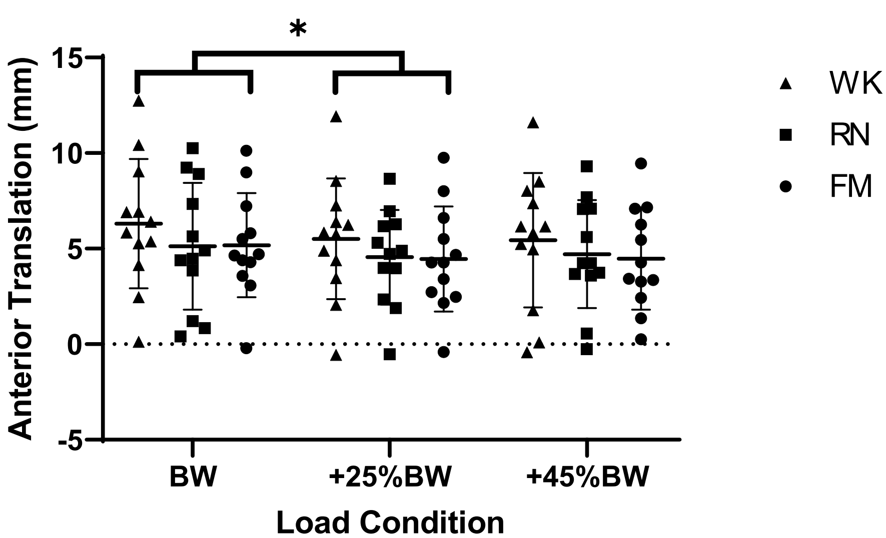

Figure 23: Average tibiofemoral anterior translation at right leg support, pairwise comparisons for velocity.

Anterior translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Tibiofemoral Kinematics at 10% Right Leg Support

Abduction

There was no significant interaction between load and locomotion in their effect on abduction ($F_{4,44} = 0.811$, $p = 0.525$, $\eta_p^2 = 0.069$). There was a significant main effect of locomotion on abduction averaged across levels of load ($F_{2,22} = 4.221$, $p = 0.028$, $\eta_p^2 = 0.277$) The level of abduction was significantly lower at the running category of locomotion (-4.4±3.5°) than the forced march category of locomotion (-3.5±3.8°), averaged across levels of load (p=0.013). There was no significant main effect of load on abduction, averaged across locomotion.

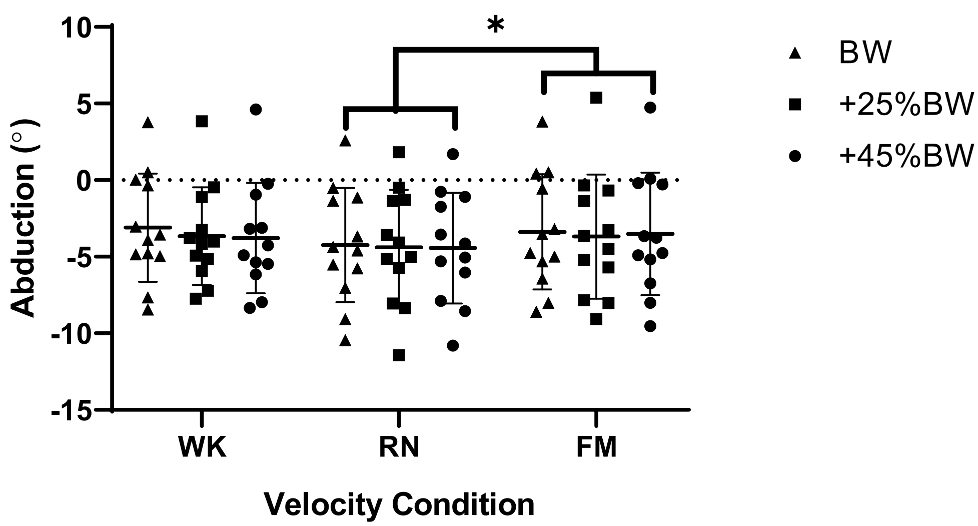

Figure 24: Average tibiofemoral abduction at 10% right leg support.

Abduction (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Internal Rotation

There was no significant interaction between load and locomotion in their effect on internal rotation ($F_{4,44} = 0.909$, $p = 0.467$, $\eta_p^2 = 0.076$). There was a significant main effect of locomotion on internal rotation averaged across levels of load ($F_{2,22} = 4.301$, $p = 0.013$, $\eta_p^2 = 0.325$). There were no significant pairwise comparisons. There was no significant main effect of load on internal rotation, averaged across locomotion.

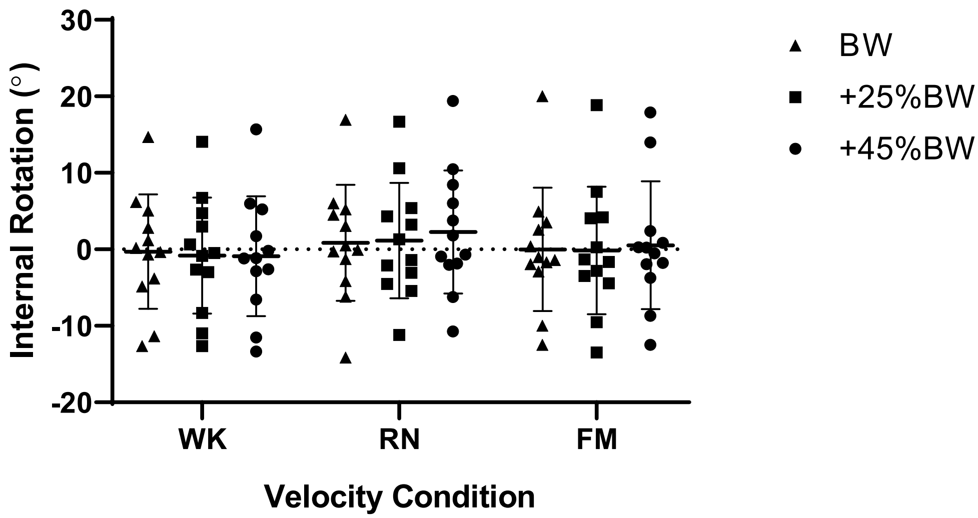

Figure 25: Average tibiofemoral internal rotation at 10% right leg support.

External rotation (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Medial Translation

There was no significant interaction between load and locomotion, in their effect on medial translation ($F_{4,44} = 1.667$, $p = 0.175$, $\eta_p^2 = 0.132$). There was no significant main effect of load on medial translation, averaged across locomotion. There was no significant main effect of locomotion on media translation, averaged across load.

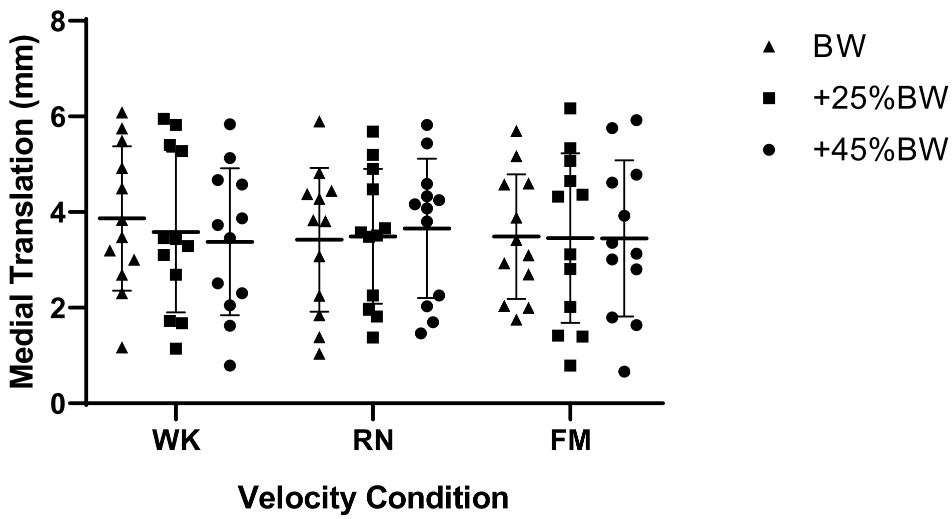

Figure 26: Average tibiofemoral medial translation at 10% right leg support.

Medial translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Proximal Translation

There was no significant interaction between load and locomotion in their effect on proximal translation ($F_{4,44} = 0.794$, $p = 0.536$, $\eta_p^2 = 0.067$). There was a significant main effect of locomotion on proximal translation averaged across levels of load ($F_{2,22} = 4.301$, $p = 0.000$, $\eta_p^2 = 0.588$). There were no significant pairwise comparisons. There was no significant main effect of load on proximal translation, averaged across locomotion.

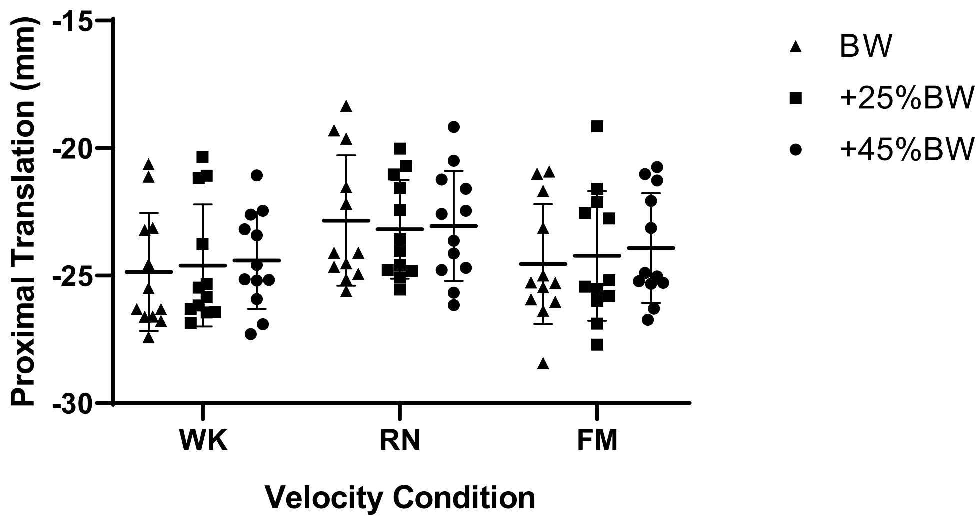

Figure 27: Average tibiofemoral proximal translation at 10% right leg support.

Proximal translation (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Anterior Translation

There was no significant interaction between load and locomotion in their effect on anterior translation ($F_{4,44} = 0.291$, $p = 0.882$, $\eta_p^2 = 0.026$). There was a significant main effect of locomotion on anterior translation averaged across levels of load ($F_{2,22} = 7.224$, $p = 0.004$, $\eta_p^2 = 0.396$). There were no significant pairwise comparisons. There was no significant main effect of load on anterior translation, averaged across locomotion.

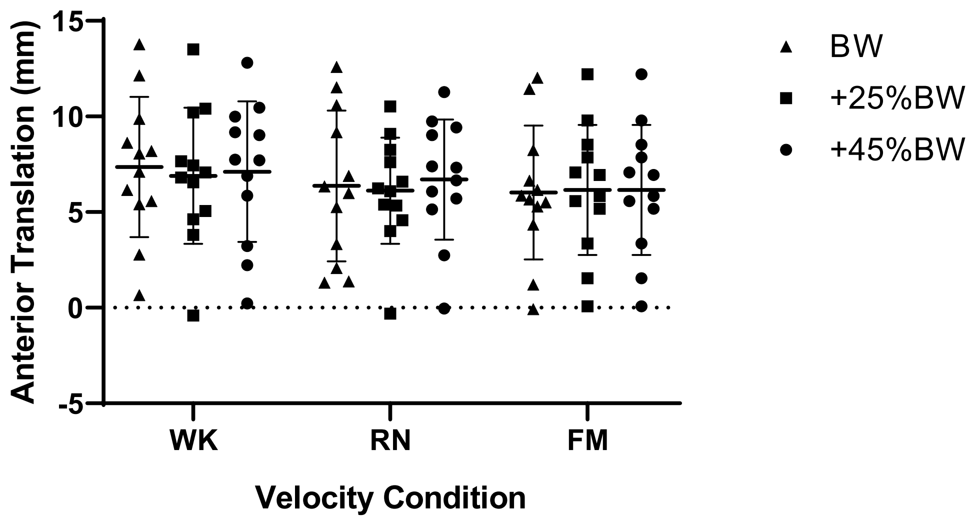

Figure 28: Average tibiofemoral anterior translation at 10% right leg support.

Anterior translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Tibiofemoral Kinematics at 20% Right Leg Support

Abduction

There was no significant interaction between load and locomotion in their effect on abduction ($F_{4,44}$ = 1.028, p = 0.403, η_p^2 = 0.085). There was no significant main effect of load on abduction, averaged across locomotion. There was no significant main effect of locomotion on abduction, averaged across load.

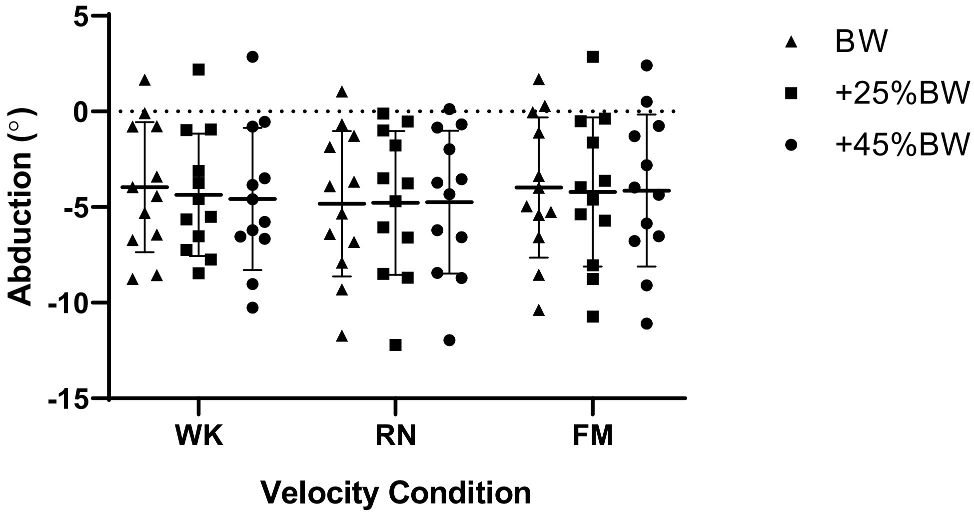

Figure 29: Average tibiofemoral abduction at 20% right leg support.

Abduction (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Internal Rotation

There was no significant interaction between load and locomotion, in their effect on internal rotation ($F_{4,44}$ = 1.742, p = 0.158, η_p^2 = 0.137). There was a significant main effect of locomotion on internal rotation, averaged across levels of load ($F_{2,22}$ = 7.081, p = 0.004, η_p^2 = 0.392). The level of internal rotation was significantly lower at the running category of locomotion (4.4±6.7°) than the walking category of locomotion (1.9±6.8°), averaged across levels of load (p = 0.036). There was no significant main effect of load on internal rotation, averaged across locomotion.

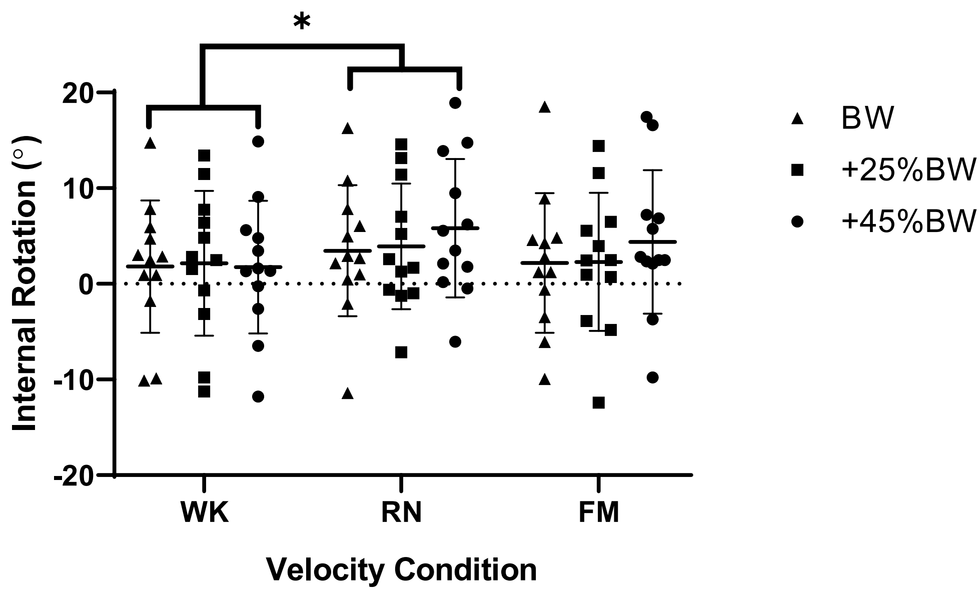

Figure 30: Average tibiofemoral internal rotation at 20% right leg support.

External rotation (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Medial Translation

There was no significant interaction between load and locomotion in their effect on medial translation ($F_{4,44} = 2.011$, $p = 0.109$, $\eta_p^2 = 0.155$). There was no significant main effect of load on medial translation, averaged across locomotion. There was no significant main effect of locomotion on medial translation, averaged across load.

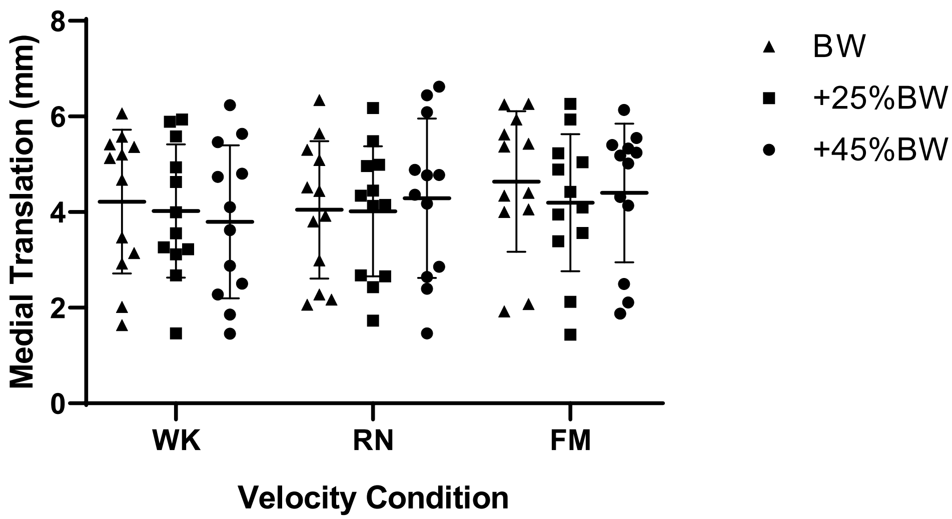

Figure 31: Average tibiofemoral medial translation at 20% right leg support.

Medial translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Proximal Translation

There was no significant interaction between load and locomotion, in their effect on proximal translation ($F_{4,44}$ = 1.110, p = 0.364, η_p^2 = 0.092). There was a significant main effect of locomotion on proximal translation, averaged across levels of load ($F_{2,22}$ = 15.497, p = 0.000, η_p^2 = 0.5853). The level of proximal translation was significantly lower at the walking level of locomotion (-22.9±2.3mm) than the running level of locomotion (-21.5±2.1 mm), averaged across levels of load (p = 0.006). There was no significant main effect of load on proximal translation, averaged across locomotion.

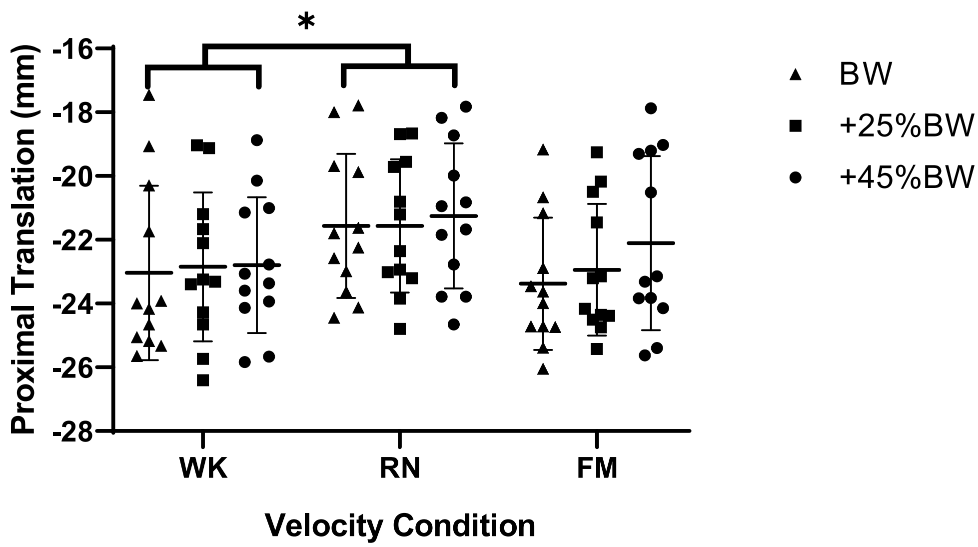

Figure 32: Average tibiofemoral proximal translation at 20% right leg support.

Proximal translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Anterior Translation

There was no significant interaction between load and locomotion, in their effect on anterior translation ($F_{4,44} = 1.327$, $p = 0.275$, $\eta_p^2 = 0.108$). There was a significant main effect of load on anterior translation, averaged across levels of locomotion ($F_{2,22} = 4.583$, $p = 0.022$, $\eta_p^2 = 0.294$). The level of anterior translation was significantly lower at the bodyweight loading condition (8.5±3.9 mm) than the +45%BW loading condition (9.4±3.7 mm), averaged across levels of locomotion (p = 0.034). There was no significant main effect of locomotion on anterior translation, averaged across load.

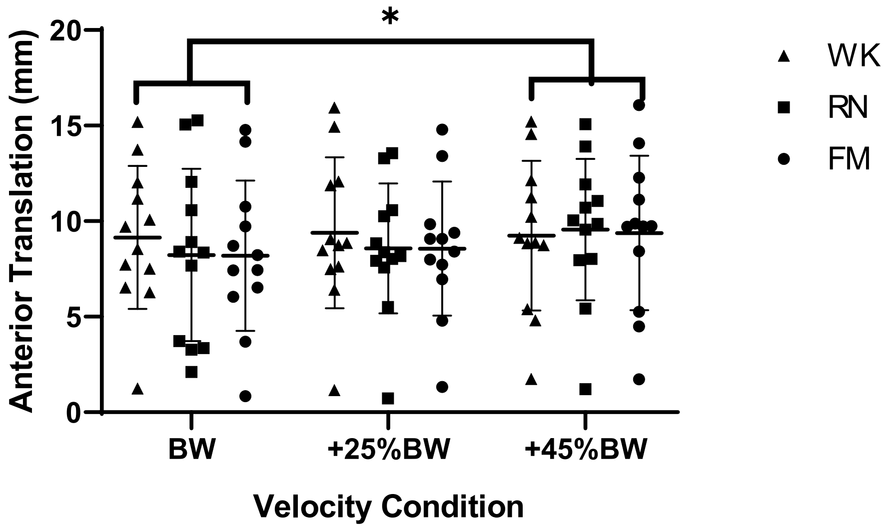

Figure 33: Average tibiofemoral anterior translation at 20% right leg support.

Anterior translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Tibiofemoral Kinematics at 30% Right Leg Support

Abduction

There was no significant interaction between load and locomotion in their effect on abduction ($F_{1.798,19.782} = 1.885$, $p = 0.181$, $\eta_p^2 = 0.146$). There was no significant main effect of load on abduction, averaged across locomotion. There was no significant main effect of locomotion on abduction, averaged across load.

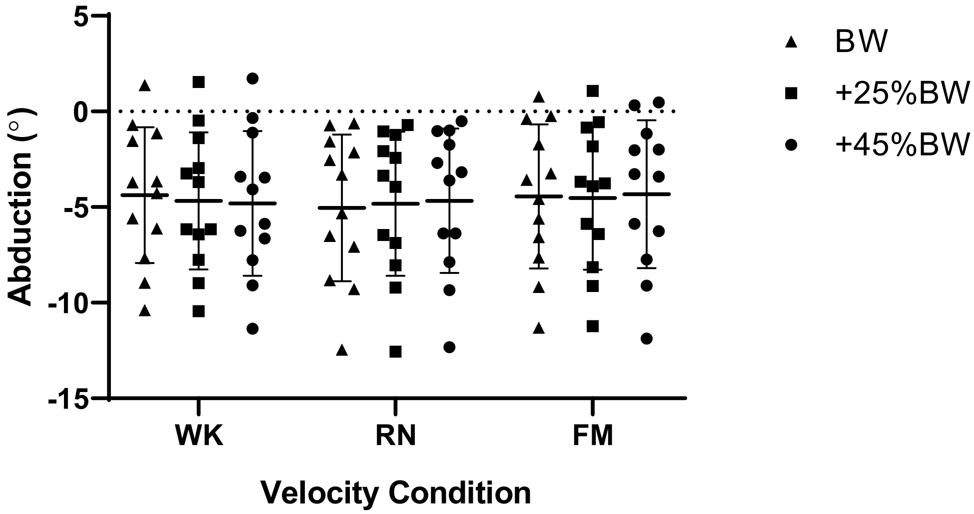

Figure 34: Average tibiofemoral abduction at 30% right leg support.

Abduction (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Internal Rotation

There was no significant interaction between load and locomotion, in their effect on internal rotation ($F_{4,44}$ = 2.000, p = 0.111, η_p^2 = 0.154). There was a significant main effect of locomotion on internal rotation, averaged across levels of load ($F_{1.206,13.269}$ = 16.088, p = 0.001, η_p^2 = 0.594). The level of internal rotation was significantly lower at the forced marching category of locomotion (5.7±6.1°) than the walking category of locomotion (3.8±6.7°, averaged across levels of load (p = 0.003). There was no significant main effect of load on internal rotation, averaged across locomotion.

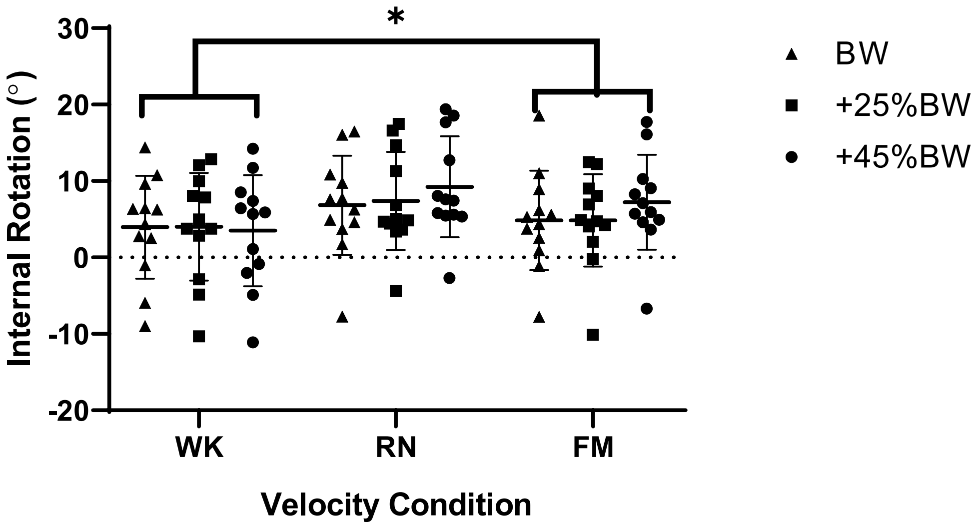

Figure 35: Average tibiofemoral internal rotation at 30% right leg support.

External rotation (°) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW)

Medial Translation

There was no significant interaction between load and locomotion, in their effect on medial translation ($F_{4,44} = 0.650$, $p = 0.630$, $\eta_p^2 = 0.056$). There was a significant main effect of locomotion on medial translation, averaged across levels of load ($F_{2,22} = 5.160$, $p = 0.015$, $\eta_p^2 = 0.319$). There were no significant pairwise comparisons. There was no significant main effect of load on medial translation, averaged across locomotion.

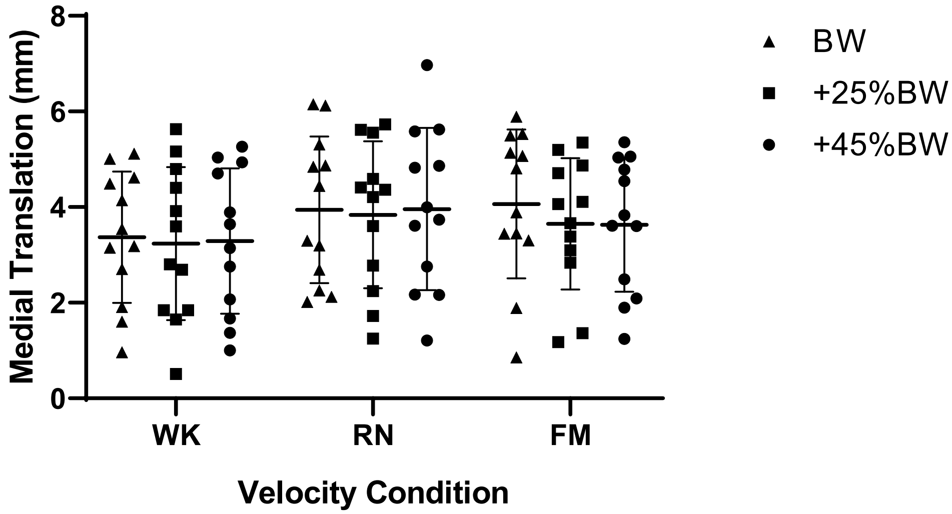

Figure 36: Average tibiofemoral medial translation at 30% right leg support.

Medial translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Proximal Translation

There was no significant interaction between load and locomotion, in their effect on proximal translation ($F_{4,44} = 0.983$, $p = 0.427$, $\eta_p^2 = 0.082$). There was a significant main effect of locomotion on proximal translation, averaged across levels of load ($F_{1.365,15.020} = 27.264$, $p = 0.000$, $\eta_p^2 = 0.713$). The level of proximal translation was significantly lower at the walking level of locomotion (-22.1±2.2 mm) than the running level of locomotion (-20.3± 2.2 mm), averaged across levels of load ($p < 0.001$). The level of proximal translation was significantly lower at the forced march level of locomotion (-21.4±2.3 mm) than the running level of locomotion (-20.3±2.2 mm), averaged across levels of load ($p < 0.001$). There was no significant main effect of load on proximal translation, averaged across locomotion.

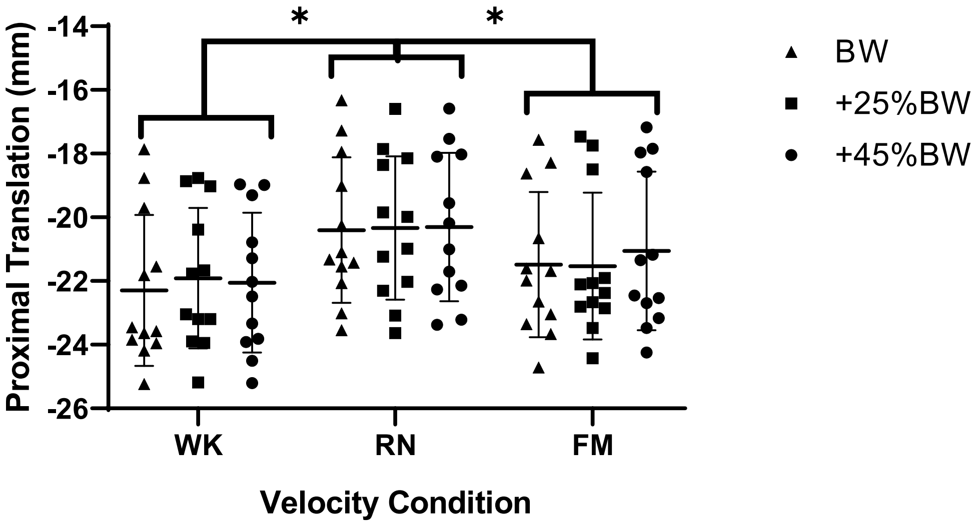

Figure 37: Average tibiofemoral proximal translation at 30% right leg support.

Proximal translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

Anterior Translation

There was no significant interaction between load and locomotion in their effect on anterior translation (4,44) = 0.971, p = 0.433, η_p^2 = 0.081). There was no significant main effect of load on anterior translation, averaged across locomotion. There was no significant main effect of locomotion on anterior translation, averaged across load.

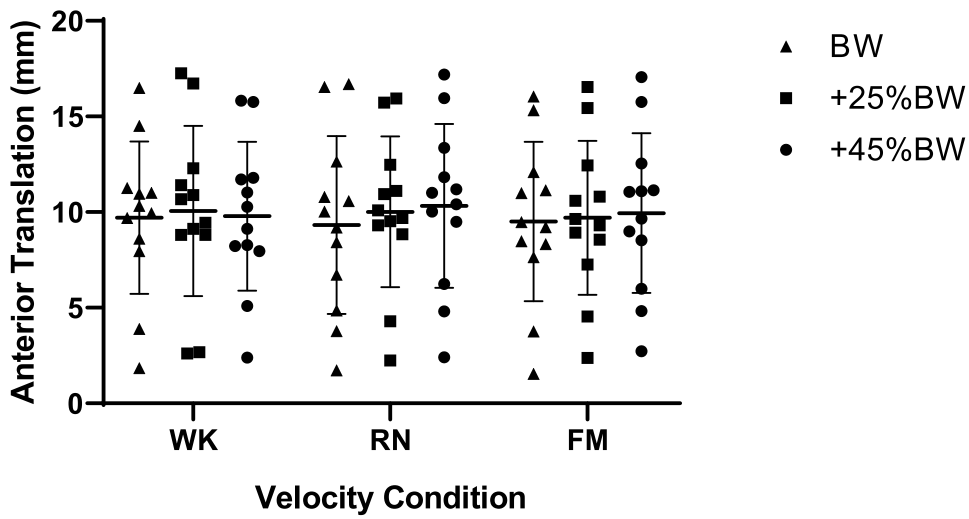

Figure 38: Average tibiofemoral anterior translation at 30% right leg support.

Anterior translation (mm) with mean ± standard deviation is shown for each velocity condition (WK, RN, FM) and load condition (BW, +25%BW, +45%BW).

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